WEST Search History

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DATE: Wednesday, March 29, 2006

Hide?	Set Name	Query	Hit Count
		DB=PGPB,USPT,USOC,EPAB,JPAB,DWPI,TDBD; PLUR=YES; OP=ADJ	
☐	L41	L31 and (((correct$4 or compensat$4) with (acquir$4 or obtain$4 or collect$4 or gather$4) with (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw) with (data or echo or signal)) with ((((amplitude or intensity or magnitude or peek or peak) with (modulation or modulating or modulate or modify or modifying or modified or chang$4 or adjust$4 or alter$3 or vary or varied or varying or variable)) or "AM")))	4
☐	L40	L32 and (((correct$4 or compensat$4) with (acquir$4 or obtain$4 or collect$4 or gather$4) with (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw) with (data or echo or signal)) with ((((amplitude or intensity or magnitude or peek or peak) with (modulation or modulating or modulate or modify or modifying or modified or chang$4 or adjust$4 or alter$3 or vary or varied or varying or variable)) or "AM")))	0
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☐	L38	L37 and (((correct$4 or compensat$4) with (acquir$4 or obtain$4 or collect$4 or gather$4) with (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw) with (data or echo or signal)) with ((((amplitude or intensity or magnitude or peek or peak) with (modulation or modulating or modulate or modify or modifying or modified or chang$4 or adjust$4 or alter$3 or vary or varied or varying or variable)) or "AM") with ((fast with spin with echo) or fse)))	0
☐	L37	L36 and ((correct$4 or compensat$4) with (acquir$4 or obtain$4 or collect$4 or gather$4) with (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw) with (data or echo or signal))	2
☐	L36	L35 and ((correct$4 or compensat$4) with (acquir$4 or obtain$4 or collect$4 or gather$4) with (data or echo or signal))	3
☐	L35	L34 and (acquir$4 or obtain$4 or collect$4 or gather$4)	8
☐	L34	L33 and (correct$4 or compensat$4)	8
☐	L33	L32 and (Fourier or "ft" or "FFT")	9
☐	L32	L31 and ((((amplitude or intensity or magnitude or peek or peak) with (modulation or modulating or modulate or modify or modifying or modified or chang$4 or adjust$4 or alter$3 or vary or varied or varying or variable)) or "AM") with ((fast with spin with echo) or fse))	9

☐	L31	L30 and ((fast with spin with echo) or fse)	290
☐	L30	L29 and (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw)	709098
☐	L29	(((amplitude or intensity or magnitude or peek or peak) with (modulation or modulating or modulate or modify or modifying or modified or chang$4 or adjust$4 or alter$3 or vary or varied or varying or variable)) or "AM")	1733527
☐	L28	l27 and (((amplitude with (modilation or modifying or modified or chang$4 or adjust$4 or alter$3 or vary or varied or varying or variable)) or "AM") with ((fast with spin with echo) or fse))	5
☐	L27	L26 and ((fast with spin with echo) or fse)	161
☐	L26	L25 and (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw)	689669
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☐	L23	L22 and (phase with (modulation or modifying or modified or chang$4 or adjust$4 or alter$3 or modulat$4 or correct$4 or compensat$4))	11
☐	L22	L21 and (phase with encod$3)	12
☐	L21	L18 and ((((amplitude) with (modulation or modifying or modified or chang$4 or adjust$4 or alter$3 or modulat$4 or correct$4 or compensat$4) or "AM")) with (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw))	16
☐	L20	L18 and ((((amplitude) with (modulation or modifying or modified or chang$4 or adjust$4 or alter$3 or modulat$4 or correct$4 or compensat$4) or "AM")) with (table ot matrix))	0
☐	L19	L18 and ((((amplitude) with (modulation or modifying or modified or chang$4 or adjust$4 or alter$3 or modulat$4 or correct$4 or compensat$4) or "AM")) with ((fast with spin with echo) or fse))	5
☐	L18	L17 and (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw)	169
☐	L17	L16 and (((amplitude) with (modulation or modifying or modified or chang$4 or adjust$4 or alter$3 or modulat$4 or correct$4 or compensat$4) or "AM"))	376
☐	L16	L15 and (amplitude)	598
☐	L15	((fast with spin with echo) or fse)	2957
☐	L14	L13 and (phase with encod$3)	1
☐	L13	(6265873)![pn]	2
☐	L12	L11 and (correct$4 or compensat$4)	5
☐	L11	L10 and (Fourier or "ft" or "FFT")	5
☐	L10	L9 and (((amplitude with (modulation or modifying or modified or chang$4 or adjust$4 or alter$3 or modulat$4)) or "AM") with ((fast with spin with echo) or fse))	5
☐	L9	L8 and (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw)	194

☐	L8	L7 and ((fast with spin with echo) or fse)	664
☐	L7	((amplitude with (modulation or modifying or modified or chang$4 or adjust$4 or alter$3 or modulat$4)) or "AM")	1373602
☐	L6	L5 and (correct$4 or compensat$4)	3
☐	L5	L4 and (Fourier or "ft" or "FFT")	3
☐	L4	L3 and (((amplitude with (modilation or modifying or modified or chang$4 or adjust$4 or alter$3)) or "AM") with((fast with spin with echo) or fse))	3
☐	L3	L2 and (k-space or kspace or "k space" or "kx" or "ky" or "kz" or "k.sub.x" or "k.sub.y" or "k.sub.z" or raw)	126
☐	L2	L1 and ((fast with spin with echo) or fse)	545
☐	L1	((amplitude with (modilation or modifying or modified or chang$4 or adjust$4 or alter$3)) or "AM")	1315469

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☐ 1. Document ID: US 6456071 B1

Using default format because multiple data bases are involved.

L6: Entry 1 of 3 File: USPT Sep 24, 2002

US-PAT-NO: 6456071
DOCUMENT-IDENTIFIER: US 6456071 B1

TITLE: Method of measuring the magnetic resonance (=NMR) by means of spin echos

DATE-ISSUED: September 24, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hennig; Jurgen	Freiburg			DE

US-CL-CURRENT: 324/307; 324/309, 324/311

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 2. Document ID: US 5345176 A

L6: Entry 2 of 3 File: USPT Sep 6, 1994

US-PAT-NO: 5345176
DOCUMENT-IDENTIFIER: US 5345176 A

TITLE: Stabilized fast spin echo NMR pulse sequence with improved slice selection

DATE-ISSUED: September 6, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
LeRoux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 3. Document ID: US 5315249 A

L6: Entry 3 of 3 File: USPT May 24, 1994

US-PAT-NO: 5315249
DOCUMENT-IDENTIFIER: US 5315249 A

TITLE: Stabilized fast spin echo NMR pulse sequence

DATE-ISSUED: May 24, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Le Roux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

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Term	Documents
CORRECT$4	0
CORRECT	1000508
CORRECTA	145
CORRECTAB	10
CORRECTABIE	2
CORRECTABJE	1
CORRECTABL	4
CORRECTABLE	9247
CORRECTABLO	1
CORRECTABLT	1
CORRECTABLY	391
(L5 AND (CORRECT$4 OR COMPENSAT$4)).PGPB,USPT,USOC,EPAB,JPAB,DWPI,TDBD.	3

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L6: Entry 1 of 3 File: USPT Sep 24, 2002

DOCUMENT-IDENTIFIER: US 6456071 B1
TITLE: Method of measuring the magnetic resonance (=NMR) by means of spin echos

Brief Summary Text (6):
Such a deviation can occur either through technical facts or be artificially produced, e.g. in applications on human beings for keeping the values of the radiated radio frequency energy within tolerable limits (SAR=specific absorption rate). Literature proposed a series of measures for limiting the corresponding signal losses. This includes on the one hand the so-called Carr-Purrcell-Meiboom-Gill method (D2) wherein by an appropriate displacement of the pulse phase between excitation and refocusing pulses, partial automatic compensation of the refocusing pulses is effected.

Brief Summary Text (9):
In applications of analytical NMR spectroscopy, improvements through different phase cycles such as MLEV16 or XY16 are used (D6). These serve mainly for compensating residual small errors in refocusing pulses with a flip angle of approximately 180.degree..

Brief Summary Text (67):
With this modification, the amplitude of the (2n+1)th echo can be reproduced to the completely refocused value (=1) for any .alpha..sub.1 . . . .alpha..sub.n. When using such a sequence in MR tomography corresponding to the RARE method, the contrast of the image is essentially given by the intensity of the echo which represents the center of the k space in the phase encoding direction.

Brief Summary Text (69):
In particular, for so-called multi-contrast methods wherein phase encoding is carried out such that at least the center of the k space is read several times and at different echo times, the principle according to equation [12] can be repeated several times even during an echo train such that several hyper-echos can be formed in one echo train.

Brief Summary Text (85):
The application, as described, onto measuring methods in MR imaging are merely illustrative. A large number of measuring sequences in analytical NMR--mainly multiple-dimensional Fourier spectroscopy--such as COSY, NOESY, INEPT, INADEQUATE etc.--to name only some of the current sequences, is based on a plurality of repetitions of multi-pulse sequences. With all these sequences, balanced magnetization can be achieved more rapidly through formation of a hyper-echo with subsequent flip back pulse and thus reduction of the measuring time and/or increase of the signal-to-noise ratio. If in such sequences, pulses are applied to different nuclei, formation of hyper-echos onto all nuclei concerned is advantageous.

Brief Summary Text (102):
Spin ensembles which move incoherently due to molecular diffusion experience an amplitude change due to the incoherent dephasing, which depends on the diffusion constant and will also attenuate the amplitude of the subsequent hyper-echo. Formation of the hyper-echo per se will not be influenced by diffusion.

Brief Summary Text (105):
The embodiments shown in FIGS. 8A through 8C of a modified hyper-echo sequence are again exemplarily. Literature (see e.g. (D9), (D10)) shows a large number of method steps which include concrete change of the signal phase and/or amplitude and can be applied also in a hyper-echo sequence.

Detailed Description Text (19):
FIG. 8C shows that a motion-dependent change of the signal phase and thus change of the amplitude of the hyper-echo can be effected already merely through corresponding magnetic field gradients alone in an otherwise unchanged hyper-echo sequence.

Detailed Description Text (22):
LITERATURE (D1) Hahn E L, Spin Echoes, Phys.Rev. 80:580-594 (1950) (D2) Meiboom S, Gill D, Modified Spin-Echo Method for Measuring Nuclear Relaxation Times, Review of Scientific Instruments, 29:688-691 (1958) (D3) Hennig J, Multiecho Imaging Sequences with Low Refocusing Flip Angles, J.Magn.Reson., 78:397-407 (1988) (D4) Le Roux P, Hinks R S, Stabilization of echo amplitudes in FSE sequences, Magn Reson Med. 30:183-90 (1993) (D5) Alsop D C, The sensitivity of low flip angle RARE imaging, Magn Reson Med. 37:176-84 (1997) (D6) Gullion T, Baker D E, Conradi M S., J.Magn.Reson. 89, 479 (1990) (D7) van Uijen C M, den Boef J H, Driven-equilibrium radiofrequency pulses in NMR imaging, Magn Reson Med. 1984 Dec;1(4):502-7. (D8) Hennig J, Thiel T, Speck O, Improved Sensitivity to Overlapping Multiplet Signals in in vivo Proton Spectroscopy Using a Multiecho Volume Selective (CPRESS-) Experiment, Magn Reson Med. 37: 816-20 (1997) (D9) Haase A, Snapshot FLASH MRI. Applications to T1, T2, and chemical-shift imaging, Magn Reson Med. 13:77-89 (1990) (D10) Norris D G, Ultrafast low-angle RARE: U-FLARE, Magn Reson Med. 17: 539-542 (1991)

Other Reference Publication (4):
Le Roux P. Hinks RS, Stabilization of echo amplitudes in FSE sequences, Magn Reson Med. 30:183-90 (1993).

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L6: Entry 2 of 3 File: USPT Sep 6, 1994

DOCUMENT-IDENTIFIER: US 5345176 A
TITLE: Stabilized fast spin echo NMR pulse sequence with improved slice selection

Abstract Text (1):
A fast spin echo NMR pulse sequence is modified to stabilize the magnitude of early NMR echo signals produced during each shot. Stabilization is accomplished by modifying the amplitude of the nutation angle produced in the spins by the corresponding RF refocusing pulses. This stabilization is relaxed for the first RF refocusing pulse so that its maximum RF power can be reduced relative to that of the other RF refocusing pulses and the bandwidth thereof increased to improve the slice select profile.

Parent Case Text (2):
This is a continuation-in-part of co-pending U.S. patent application Ser. No. 07/920,952, filed on Jul. 28, 1992 and entitled "Stabilized Fast Spin Echo NMR Pulse Sequence" now U.S. Pat. No. 5,315,249.

Brief Summary Text (2):
The field of the invention is nuclear magnetic resonance imaging methods and systems. More particularly, the invention relates to the reduction of image artifacts in fast spin-echo (FSE) pulse sequences by producing RF refocusing pulses which stabilize the magnitude of the acquired spin echo signals.

Brief Summary Text (10):
Both of these "fast spin echo" imaging methods involve the acquisition of multiple spin echo signals from a single excitation pulse in which each acquired echo signal is separately phase encoded. Each pulse sequence, or "shot," therefore results in the acquisition of a plurality of views and a plurality of shots are typically employed to acquire a complete set of image data. For example, a RARE pulse sequence might acquire 8 or 16 separate echo signals, per shot, and an image requiring 256 views would, therefore, require 32 or 16 shots respectively.

Brief Summary Text (11):
It is well known that the RARE sequence, and particularly its slice selective implementation, suffers from a non-steady state behavior in the first NMR echo signals acquired during each shot. This is particularly true when the selective RF refocusing pulses are not exactly 180.degree.. In our copending U.S. patent application Ser. No. 07/920,952, filed on Jul. 28, 1992 and entitled "Stablized Fast Spin Echo NMR Pulse Sequence" we describe a technique for altering the nutation angles in successive RF refocusing pulses in order to stabilize the early NMR echo signals acquired during the RARE sequence. When applied to selective RF refocusing pulses this not only affects their amplitude, but also their shape. As a result, to provide perfectly stablized signals very large RF excitation fields must be produced if the proper shape of the selected slice is to be maintained. Particularly when the first spin echo signal is fully stabilized in a three-dimensional acquisition, the desired thickness and shape of the slice cannot be achieved with the RF power available on commercial NMR systems.

Brief Summary Text (13):
The present invention relates to an improved fast spin echo pulse sequence in which

the magnitude of an acquired series of NMR spin-echo signals is stabilized by shaping the RF refocusing pulses which produce them. More particularly, in a fast spin echo pulse sequence one or more RF refocusing pulses in a series are modified such that the magnitudes of their corresponding NMR spin-echo signals do not significantly oscillate, and the first RF refocusing pulse in the series is further modified to reduce the magnitude of its corresponding NMR spin-echo signal a predetermined amount. By modifying the first RF refocusing pulse it can be produced with the bandwidth and RF power needed to provide the desired slice, or slab, selection; and because the magnitude of the corresponding NMR spin-echo signal is reduced by a predetermined amount, it can be properly compensated in the image reconstruction process.

Drawing Description Text (4):
FIG. 3 is a graphic representation of a fast spin-echo pulse sequence;

Drawing Description Text (6):
FIG. 5 is a graphic representation of the RF refocusing pulse magnitude required for each RF refocusing pulse in the FSE pulse sequence of FIG. 3 to provide stabilized NMR echo signals according to the present invention; and

Detailed Description Text (2):
Referring first to FIG. 1, there is shown in block diagram form the major components of a preferred NMR system which incorporates the present invention and which is sold by the General Electric Company under the trademark "SIGNA". The overall operation of the system is under the control of a host computer system generally designated 100 which includes a main computer 101 (such as a Data General MV7800). The computer has associated therewith an interface 102 through which a plurality of computer peripheral devices and other NMR system components are coupled. Among the computer peripheral devices is a magnetic tape drive 104 which may be utilized under the direction of the main computer for archiving patient data and images to tape. Processed patient data may also be stored in an image disc storage device designated 110. The function of image processor 108 is to provide interactive image display manipulation such as magnification, image comparison, gray-scale adjustment and real-time data display. The computer system is provided with a means to store raw data (i.e. before image construction) utilizing a disc data storage system designated 112. An operator console 116 is also coupled to the computer by means of interface 102 and provides the operator with the means to input data pertinent to a patient study as well as additional data necessary for proper NMR system operation, such as calibrating, initiating and terminating scans. The operator console is also used to display images stored on discs or magnetic tape.

Detailed Description Text (6):
The gradient coil assembly 136 and the RF transmit and receiver coils 138 are mounted within the bore of the magnet utilized to produce the polarizing magnetic field. The magnet forms a part of the main magnet assembly which includes the patient alignment system 148. A shim power supply 140 is utilized to energize a shim coil associated with the main magnet and which are used to correct inhomogeneities in the polarizing magnet field. In the case of a superconductive magnet, the main power supply 142 is utilized to bring the polarizing field produced by the magnet to the proper operating strength and is then disconnected. The patient alignment system 148 operates in combination with a patient cradle and transport system 150 and patient positioning system 152. To minimize interference from external sources, these NMR system components are enclosed in an RF-shielded room generally designated 144.

Detailed Description Text (11):
Referring particularly to FIG. 3, a conventional fast spin echo NMR pulse sequence, known as a 2DFT RARE sequence is shown. For clarity, only four echo signals 301-304 are shown in FIG. 3, but it can be appreciated that more are produced and acquired.

These NMR echo signals are produced by a 90.degree. RF excitation pulse 305 which is generated in the presence of a G.sub.z slice select gradient pulse 306 to provide transverse magnetization in a slice through the patient. This transverse magnetization is refocused by each selective RF refocusing pulse 307 to produce the NMR spin echo signals 301-304 that are acquired in the presence of G.sub.x readout gradient pulses 308. Each NMR spin echo signal 301-304 is separately phase encoded by respective G.sub.y phase encoding pulses 309-313. The magnitude of each phase encoding pulse is different, and it is stepped through 256 values to acquire 256 separate views during a complete scan. This enables an image having 256 separate pixels in the y direction to be reconstructed. Each NMR spin echo signal is acquired by digitizing 256 samples of each signal. As a result, at the completion of a scan for one image, 16 shots (256/16=16) of the pulse sequence of FIG. 3 have been executed and a 256 by 256 element array of complex numbers has been acquired An image is reconstructed by performing a 2D Fourier transformation on this image data array and then calculating the absolute value of each resulting complex element. A 256 by 256 pixel image is thus produced in which the brightness of each pixel is determined by the magnitude of its corresponding element in the transformed array.

Detailed Description Text (12):
Referring still to FIG. 3, the T2 decay in the NMR spin echo signals 301-304 is illustrated by the dashed line 315. The rate of decay is different for different tissue types and a common strategy in FSE NMR imaging is to enhance the contrast in certain tissues over other tissues by judiciously selecting an effective echo time. This effective echo time is determined primarily by the actual echo time (TE) of the central, or low-order, views that dominate image contrast. For example, to enhance muscle tissue in the image of a human knee joint, the first spin echo signals may be encoded to a low-order phase encoding value in each shot because the T2 decay rate of muscle tissue is high and the shortest possible effective echo time (TE) is desired. On the other hand, to produce an image in which the fluids in the knee joint are enhanced, the low-order phase encoding views may be acquired from later echo signals which have a much longer echo time TE. The T2 decay rate of joint fluids is much less than that of muscle tissue, and as a result, these fluids will contribute proportionately more signal and their contrast will be enhanced in comparison with that of muscle tissue.

Detailed Description Text (13):
With the conventional FSE pulse sequence, the NMR echo signals 301-304 do not decay smoothly along the dashed line 315. Instead, the magnitude of the NMR signals 301-305 may oscillate significantly below this optimal T.sub.2 decay curve 315, particularly during the early NMR echo signals. This is illustrated in FIG. 4, where T.sub.2 is assumed to be very large, the vertical axis is NMR echo signal strength, and the horizontal axis is the number of the NMR echo signal in the shot. Each line represents the magnitude of the NMR echo signals produced by RF refocusing pulses having the indicated constant tip angle. The figure illustrates tip angles from .theta.=10.degree. to .theta.=170.degree., and it should be apparent from these that the signal level variation problem does not arise when perfect 180.degree. RF refocusing pulses are produced. Instead, as the tip angle is reduced below 180.degree., the oscillations in the early NMR echo signal magnitudes become very significant even at tip angles marginally less than 180.degree.. As the tip angle is further decreased, more NMR echo signals are affected before an equilibrium condition is reached, but the oscillations become less pronounced.

Detailed Description Text (26):
When the series of RF refocusing pulses are modified to produce a substantially constant NMR spin-echo signal amplitude (S), the modifications to the first few refocusing pulses is quite substantial. When applied to selective RF refocusing pulses, the above-cited copending application teaches that the single slice selective RF refocusing pulse is disected into a set of contiguous subslices and that each subslice in each RF refocusing pulse should be compensated to produce the

desired signal amplitude (S). When this is done, however, the shape of the first selective RF refocusing pulse is changed such that it has a larger bandwidth than the other RF refocusing pulses. In consequence its maximum amplitude of the time domain RF excitation field B.sub.1 is large and reaches quickly the maximum amplitude (or peak power) that the RF transmitter can play, particularly under heavily loaded conditions. This limits the selectivity of the RF refocusing pulses rather than the classical time transition-bandwidth product. We prefer not to increase the time-duration of the RF refocusing pulses to offset this limitation, since a goal in FSE imaging is to reduce the time between NMR echo signals.

CLAIMS:

1. An NMR system, the combination comprising:

means for generating a polarizing magnetic field;

excitation means for generating an RF excitation magnetic field which produces transverse magnetization in spins subjected to the polarizing magnetic field;

receiver means for sensing an NMR signal produced by the transverse magnetization and producing digitized samples of the NMR signal;

first gradient means for generating a first magnetic field gradient to phase encode the NMR signal;

second gradient means for generating a second magnetic field gradient to frequency encode the NMR signal; and

pulse control means coupled to the excitation means, first gradient means, second gradient means, and receiver means, said pulse control means being operable to conduct a fast spin echo pulse sequence in which a series of NMR echo signals are produced in response to a corresponding series of RF refocusing pulses produced by said excitation means, and in which a set of NMR echo signals following the first NMR echo signal in said series of NMR echo signals are stabilized to have a substantially similar amplitude (S) by altering the flip angle produced by RF refocusing pulses in said series, and the flip angle (.theta.) produced by the first RF refocusing pulse in said series is set to substantially the same flip angle (.theta.) as that of the second RF refocusing pulse in said series.

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L6: Entry 3 of 3 File: USPT May 24, 1994

DOCUMENT-IDENTIFIER: US 5315249 A
TITLE: Stabilized fast spin echo NMR pulse sequence

Abstract Text (1):
A fast spin echo NMR pulse sequence is modified to stabilize the magnitude of early NMR echo signals produced during each shot. Stabilization is accomplished by modifying the amplitude of the nutation angle produced in the spins by the corresponding RF refocusing pulses. When selective RF refocusing pulses are employed the slice is divided into subslices and the modifications are made separately to each subslice.

Brief Summary Text (2):
The field of the invention is nuclear magnetic resonance imaging methods and systems. More particularly, the invention relates to the reduction of image artifacts in fast spin-echo (FSE) pulse sequences by producing RF refocusing pulses which stabilize the magnitude of the acquired spin echo signals.

Brief Summary Text (10):
Both of these "fast spin echo" imaging methods involve the acquisition of multiple spin echo signals from a single excitation pulse in which each acquired echo signal is separately phase encoded. Each pulse sequence, or "shot," therefore results in the acquisition of a plurality of views and a plurality of shots are typically employed to acquire a complete set of image data. For example, a RARE pulse sequence might acquire 8 or 16 separate echo signals, per shot, and an image requiring 256 views would, therefore, require 32 or 16 shots respectively.

Brief Summary Text (13):
The present invention relates to an improved fast spin-echo pulse sequence in which the magnitude of an acquired NMR spin echo signal is stabilized by shaping the RF refocusing pulse which produces it. More particularly, in a fast spin echo pulse sequence one or more RF refocusing pulses are modified by changing their modulation envelope such that the magnitude of the NMR spin-echo signals does not oscillate. In a selective RF refocusing pulse this is accomplished by treating the RF refocus pulse slice profile as a series of subslices which each have a different tip angle and which each must be separately compensated.

Brief Summary Text (14):
A general object of the invention is to compensate the selective RF refocusing pulses in a fast spin echo sequence such that the NMR spin echo signals are stabilized in magnitude. It has been discovered that the amount of instability in the NMR echo signals is a function of the tip angle of the refocusing pulses. At a true 180.degree. tip angle there is no instability, but as the tip angle becomes smaller, the fluctuations in NMR echo signal magnitude increase. It is one discovery of the present invention that for a given echo signal magnitude, the magnitude and phase of the RF refocusing pulses in an FSE pulse sequence can be calculated such that all the resulting NMR spin echo signals may be stabilized. It is a further discovery of the present invention that since a selective RF refocusing pulse actually produces a range of tip angles over the thickness of the

slice, then to properly stabilize the NMR spin echo signals produced by such selective RF refocusing pulses, the slice profile may be considered a set of subslices at different tip angles. Accordingly, RF refocusing pulses may be produced to achieve excitation profiles that result in the stabilization of each subslice, and thus result in stabilization of the entire slice.

Brief Summary Text (15):
A general object of the invention is to stabilize the NMR spin echo signals in an FSE pulse sequence without increasing the scan time. No additional pulses need be added to the FSE pulse sequence. Instead, the shape of the RF refocusing pulse envelope is changed on as many of the initial refocusing pulses in the sequence as is necessary to provide the desired degree of stabilization. The modified pulse shapes may be calculated and stored in advance of the scan, and are played out in real time as the scan is conducted in the same manner as unmodified RF refocusing pulses.

Drawing Description Text (4):
FIG. 3 is a graphic representation of a fast spin-echo pulse sequence;

Drawing Description Text (7):
FIG. 6 is a graphic representation of the RF refocusing pulse magnitude required for each RF refocusing pulse in the FSE pulse sequence of FIG. 3 to provide stabilized NMR echo signals according to the present invention.

Detailed Description Text (2):
Referring first to FIG. 1, there is shown in block diagram form the major components of a preferred NMR system which incorporates the present invention and which is sold by the General Electric Company under the trademark "SIGNA". The overall operation of the system is under the control of a host computer system generally designated 100 which includes a main computer 101 (such as a Data General MV7800). The computer has associated therewith an interface 102 through which a plurality of computer peripheral devices and other NMR system components are coupled. Among the computer peripheral devices is a magnetic tape drive 104 which may be utilized under the direction of the main computer for archiving patient data and images to tape. Processed patient data may also be stored in an image disc storage device designated 110. The function of image processor 108 is to provide interactive image display manipulation such as magnification, image comparison, gray-scale adjustment and real-time data display. The computer system is provided with a means to store raw data (i.e. before image construction) utilizing a disc data storage system designated 112. An operator console 116 is also coupled to the computer by means of interface 102 and provides the operator with the means to input data pertinent to a patient study as well as additional data necessary for proper NMR system operation, such as calibrating, initiating and terminating scans. The operator console is also used to display images stored on discs or magnetic tape.

Detailed Description Text (6):
The gradient coil assembly 136 and the RF transmit and receiver coils 138 are mounted within the bore of the magnet utilized to produce the polarizing magnetic field. The magnet forms a part of the main magnet assembly which includes the patient alignment system 148. A shim power supply 140 is utilized to energize a shim coil associated with the main magnet and which are used to correct inhomogeneities in the polarizing magnet field. In the case of a superconductive magnet, the main power supply 142 is utilized to bring the polarizing field produced by the magnet to the proper operating strength and is then disconnected. The patient alignment system 148 operates in combination with a patient cradle and transport system 150 and patient positioning system 152. To minimize interference from external sources, these NMR system components are enclosed in an RF-shielded room generally designated 144.

Detailed Description Text (11):
Referring particularly to FIG. 3, a conventional fast spin echo NMR pulse sequence, known as a 2DFT RARE sequence is shown. For clarity, only four echo signals 301-304 are shown in FIG. 3, but it can be appreciated that more are produced and acquired. These NMR echo signals are produced by a 90.degree. RF excitation pulse 305 which is generated in the presence of a G.sub.z slice select gradient pulse 306 to provide transverse magnetization in a slice through the patient. This transverse magnetization is refocused by each selective RF refocusing pulse 307 to produce the NMR spin echo signals 301-304 that are acquired in the presence of G.sub.x readout gradient pulses 308. Each NMR spin echo signal 301-304 is separately phase encoded by respective G.sub.y phase encoding pulses 309-313. The magnitude of each phase encoding pulse is different, and it is stepped through 256 values to acquire 256 separate views during a complete scan. This enables an image having 256 separate pixels in the y direction to be reconstructed. Each NMR spin echo signal is acquired by digitizing 256 samples of each signal. As a result, at the completion of a scan for one image, 16 shots (256/16=16) of the pulse sequence of FIG. 3 have been executed and a 256 by 256 element array of complex numbers have been acquired. An 20 image is reconstructed by performing a 2D Fourier transformation on this image data array and then calculating the absolute value of each resulting complex element. A 256 by 256 pixel image is thus produced in which the brightness of each pixel is determined by the magnitude of its corresponding element in the transformed array.

Detailed Description Text (12):
Referring still to FIG. 3, the T.sub.2 decay in the NMR spin echo signals 301-304 is illustrated by the dashed line 315. The rate of decay is different for different tissue types and a common strategy in FSE NMR imaging is to enhance the contrast in certain tissues over other tissues by judiciously selecting an effective echo time. This effective echo time is determined primarily by the actual echo time (TE) of the central, or low-order, views that dominate image contrast. For example, to enhance muscle tissue in the image of a human knee joint, the first spin echo signals may be encoded to a low-order phase encoding value in each shot because the T.sub.2 decay rate of muscle tissue is high and the shortest possible effective echo time (TE) is desired. On the other hand, to produce an image in which the fluids in the knee joint are enhanced, the low-order phase encoding views may be acquired from later echo signals which have a much longer echo time TE. The T.sub.2 decay rate of joint fluids is much less than that of muscle tissue, and as a result, these fluids will contribute proportionately more signal and their contrast will be enhanced in comparison with that of muscle tissue.

Detailed Description Text (13):
With the conventional FSE pulse sequence, the NMR echo signals 301-304 do not decay smoothly along the dashed line 315. Instead, the magnitude of the NMR signals 301-305 may oscillate significantly below this optimal T.sub.2 decay curve 315, particularly during the early NMR echo signals. This is illustrated in FIG. 4, where T.sub.2 is assumed to be very large, the vertical axis is NMR echo signal strength, and the horizontal axis is the number of the NMR echo signal in the shot. Each line represents the magnitude of the NMR echo signals produced by RF refocusing pulses having the indicated constant tip angle. The figure illustrates tip angles from .theta.= 10.degree. to .theta.= 170.degree., and it should be apparent from these that the signal level variation problem does not arise when perfect 180.degree. RF refocusing pulses are produced. Instead, as the tip angle is reduced below 180.degree., the oscillations in the early NMR echo signal magnitudes become very significant even at tip angles marginally less than 180.degree.. As the tip angle is further decreased, more NMR echo signals are affected before an equilibrium condition is reached, but the oscillations become less pronounced.

Detailed Description Text (26):
However, as indicated above, the present invention is also applicable where selective 180.degree. RF refocusing pulses are employed in the FSE sequence.

Referring again to FIG. 5, in a conventional slice select profile 319 a slab of spins over a region indicated as subslice 320 is excited at the desired 180.degree. nutation angle. However, this is not true in the transition regions 330 on each edge of this central subslice 320. Instead, these transition regions 330 can be viewed as a set of separate subslices having separate tip angles. Unless stabilized, the spins in these transition regions 330 will produce NMR signal components which vary in amplitude quite significantly during the early portion of each FSE shot. The contribution which these transition spins make to the total NMR echo signal will vary as a function of slice thickness and slice profile, however, it is a significant amount and the resulting echo signals will oscillate in magnitude.

CLAIMS:

1. An NMR system, the combination comprising:

means for generating a polarizing magnetic field;

excitation means for generating an RF excitation magnetic field which produces transverse magnetization in spins subjected to the polarizing magnetic field;

receiving means for sensing an NMR signal produced by the transverse magnetization and producing digitized samples of the NMR signal;

first gradient means for generating a first magnetic field gradient to phase encode the NMR signal;

second gradient means for generating a second magnetic field gradient to frequency encode the NMR signal; and

pulse control means coupled to the excitation means, first gradient means, second gradient means, receiver means, said pulse control means being operable to conduct a fast spin echo pulse sequence in which a series of NMR echo signals are produced in response to a single RF excitation pulse followed by a corresponding series of RF refocusing pulses produced by said excitation means, and in which said NMR echo signals are stabilized to a substantially smoothly decaying amplitude by altering the flip angle produced by one or more of the initial RF refocusing pulses in said series.

2. An NMR system, the combination comprising:

means for generating a polarizing magnetic field;

excitation means for generating an RF excitation magnetic field which produces transverse magnetization in spins subjected to the polarizing magnetic field;

receiver means for sensing an NMR signal produced by the transverse magnetization and producing digitized samples of the NMR signal;

first gradient means for generating a first magnetic gradient to phase encode the NMR signal;

second gradient means for generating a second magnetic field gradient to frequency encode the NMR signal;

third gradient means for generating a third magnetic field gradient to select a slice of said spins comprised of a plurality of adjacent subslices of said spins which are transversely magnetized by said excitation means; and

pulse control means coupled to the excitation means, first gradient means, second

gradient means, third gradient means and receiver means, said pulse control module means being operable to conduct a fast spin echo pulse sequence in which a series of NMR echo signals are produced in response to a single RF excitation pulse followed by a corresponding series of selective RF refocusing pulses produced by said excitation means concurrently with corresponding slice select pulses produced by said third gradient means, and in which said NMR echo signals are stabilized to a substantially smoothly decaying amplitude by altering the flip angle produced by said selective RF refocusing pulses for one or more subslice components.

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Hit List

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Search Results - Record(s) 1 through 5 of 5 returned.

☐ 1. Document ID: US 6456071 B1

Using default format because multiple data bases are involved.

L28: Entry 1 of 5 File: USPT Sep 24, 2002

US-PAT-NO: 6456071
DOCUMENT-IDENTIFIER: US 6456071 B1

TITLE: Method of measuring the magnetic resonance (=NMR) by means of spin echos

DATE-ISSUED: September 24, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hennig; Jurgen	Freiburg			DE

US-CL-CURRENT: 324/307; 324/309, 324/311

Full | Title | Citation | Front | Review | Classification | Date | Reference | Claims | KMC | Draw D

☐ 2. Document ID: US 5578923 A

L28: Entry 2 of 5 File: USPT Nov 26, 1996

US-PAT-NO: 5578923
DOCUMENT-IDENTIFIER: US 5578923 A

TITLE: T2 restoration and noise suppression of hybrid MR images using wiener and linear prediction techniques

DATE-ISSUED: November 26, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/309; 128/925

Full | Title | Citation | Front | Review | Classification | Date | Reference | Claims | KMC | Draw D

☐ 3. Document ID: US 5517122 A

L28: Entry 3 of 5 File: USPT May 14, 1996

US-PAT-NO: 5517122
DOCUMENT-IDENTIFIER: US 5517122 A

TITLE: T2 restoration and noise suppression of hybrid MR images using Wiener and linear prediction techniques

DATE-ISSUED: May 14, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/322; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KMC | Draw D

☐ 4. Document ID: US 5345176 A

L28: Entry 4 of 5 File: USPT Sep 6, 1994

US-PAT-NO: 5345176
DOCUMENT-IDENTIFIER: US 5345176 A

TITLE: Stabilized fast spin echo NMR pulse sequence with improved slice selection

DATE-ISSUED: September 6, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
LeRoux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KMC | Draw D

☐ 5. Document ID: US 5315249 A

L28: Entry 5 of 5 File: USPT May 24, 1994

US-PAT-NO: 5315249
DOCUMENT-IDENTIFIER: US 5315249 A

TITLE: Stabilized fast spin echo NMR pulse sequence

DATE-ISSUED: May 24, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Le Roux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw De

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs | Generate OACS

Term	Documents
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AMPLITUDES	107302
MODILATION	122
MODILATIONS	4
MODIFYING	371799
MODIFYINGS	4
MODIFIED	1764797
MODIFIEDS	17
VARY	1556637
VARIES	775775
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Search Results - Record(s) 1 through 9 of 9 returned.

☐ 1. Document ID: US 6700375 B2

Using default format because multiple data bases are involved.

L33: Entry 1 of 9 File: USPT Mar 2, 2004

US-PAT-NO: 6700375
DOCUMENT-IDENTIFIER: US 6700375 B2

TITLE: MRI method and apparatus using independent gradient magnetic field spolier pulses

DATE-ISSUED: March 2, 2004

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Machida; Yoshio	Nasu-gun			JP
Sugiura; Satoshi	Otawara			JP
Kassai; Yoshimori	Nasu-gun			JP

US-CL-CURRENT: 324/314; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 2. Document ID: US 6456071 B1

L33: Entry 2 of 9 File: USPT Sep 24, 2002

US-PAT-NO: 6456071
DOCUMENT-IDENTIFIER: US 6456071 B1

TITLE: Method of measuring the magnetic resonance (=NMR) by means of spin echos

DATE-ISSUED: September 24, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hennig; Jurgen	Freiburg			DE

US-CL-CURRENT: 324/307; 324/309, 324/311

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

3. Document ID: US 6043654 A

L33: Entry 3 of 9 File: USPT Mar 28, 2000

US-PAT-NO: 6043654
DOCUMENT-IDENTIFIER: US 6043654 A

TITLE: Multi-volume slicing and interleaved phase-encoding acquisition for 3 D fast spin echo (FSE)

DATE-ISSUED: March 28, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Liu; Kecheng	Solon	OH		
Xu; Yansun	Solon	OH		
Loncar; Mark J.	Richmond Heights	OH		

US-CL-CURRENT: 324/309; 324/306

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw De

4. Document ID: US 5779636 A

L33: Entry 4 of 9 File: USPT Jul 14, 1998

US-PAT-NO: 5779636
DOCUMENT-IDENTIFIER: US 5779636 A

TITLE: Method of echo volume imaging and MRI system using the same

DATE-ISSUED: July 14, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Kanazawa; Hitoshi	Nasu-Gun			JP

US-CL-CURRENT: 600/410; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw De

5. Document ID: US 5578923 A

L33: Entry 5 of 9 File: USPT Nov 26, 1996

US-PAT-NO: 5578923
DOCUMENT-IDENTIFIER: US 5578923 A

TITLE: T2 restoration and noise suppression of hybrid MR images using wiener and linear prediction techniques

DATE-ISSUED: November 26, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/309; 128/925

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 6. Document ID: US 5517122 A

L33: Entry 6 of 9 File: USPT May 14, 1996

US-PAT-NO: 5517122
DOCUMENT-IDENTIFIER: US 5517122 A

TITLE: T2 restoration and noise suppression of hybrid MR images using Wiener and linear prediction techniques

DATE-ISSUED: May 14, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/322; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 7. Document ID: US 5345176 A

L33: Entry 7 of 9 File: USPT Sep 6, 1994

US-PAT-NO: 5345176
DOCUMENT-IDENTIFIER: US 5345176 A

TITLE: Stabilized fast spin echo NMR pulse sequence with improved slice selection

DATE-ISSUED: September 6, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
LeRoux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 8. Document ID: US 5315249 A

L33: Entry 8 of 9 File: USPT May 24, 1994

US-PAT-NO: 5315249
DOCUMENT-IDENTIFIER: US 5315249 A

TITLE: Stabilized fast spin echo NMR pulse sequence

DATE-ISSUED: May 24, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Le Roux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 9. Document ID: US 4684891 A

L33: Entry 9 of 9 File: USPT Aug. 4, 1987

US-PAT-NO: 4684891
DOCUMENT-IDENTIFIER: US 4684891 A

TITLE: Rapid magnetic resonance imaging using multiple phase encoded spin echoes in each of plural measurement cycles

DATE-ISSUED: August 4, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Feinberg; David A.	Berkeley	CA		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

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Term	Documents
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FT	377332
FTS	30805

FFT	25897
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Search Results - Record(s) 1 through 8 of 8 returned.

☐ 1. Document ID: US 6456071 B1

Using default format because multiple data bases are involved.

L34: Entry 1 of 8 File: USPT Sep 24, 2002

US-PAT-NO: 6456071
DOCUMENT-IDENTIFIER: US 6456071 B1

TITLE: Method of measuring the magnetic resonance (=NMR) by means of spin echos

DATE-ISSUED: September 24, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hennig; Jurgen	Freiburg			DE

US-CL-CURRENT: 324/307; 324/309, 324/311

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 2. Document ID: US 6043654 A

L34: Entry 2 of 8 File: USPT Mar 28, 2000

US-PAT-NO: 6043654
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DATE-ISSUED: March 28, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Liu; Kecheng	Solon	OH		
Xu; Yansun	Solon	OH		
Loncar; Mark J.	Richmond Heights	OH		

US-CL-CURRENT: 324/309; 324/306

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 3. Document ID: US 5779636 A

L34: Entry 3 of 8 File: USPT Jul 14, 1998

US-PAT-NO: 5779636
DOCUMENT-IDENTIFIER: US 5779636 A

TITLE: Method of echo volume imaging and MRI system using the same

DATE-ISSUED: July 14, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Kanazawa; Hitoshi	Nasu-Gun			JP

US-CL-CURRENT: 600/410; 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 4. Document ID: US 5578923 A

L34: Entry 4 of 8 File: USPT Nov 26, 1996

US-PAT-NO: 5578923
DOCUMENT-IDENTIFIER: US 5578923 A

TITLE: T2 restoration and noise suppression of hybrid MR images using wiener and linear prediction techniques

DATE-ISSUED: November 26, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/309; 128/925

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 5. Document ID: US 5517122 A

L34: Entry 5 of 8 File: USPT May 14, 1996

US-PAT-NO: 5517122
DOCUMENT-IDENTIFIER: US 5517122 A

TITLE: T2 restoration and noise suppression of hybrid MR images using Wiener and linear prediction techniques

DATE-ISSUED: May 14, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/322; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 6. Document ID: US 5345176 A

L34: Entry 6 of 8 File: USPT Sep 6, 1994

US-PAT-NO: 5345176
DOCUMENT-IDENTIFIER: US 5345176 A

TITLE: Stabilized fast spin echo NMR pulse sequence with improved slice selection

DATE-ISSUED: September 6, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
LeRoux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 7. Document ID: US 5315249 A

L34: Entry 7 of 8 File: USPT May 24, 1994

US-PAT-NO: 5315249
DOCUMENT-IDENTIFIER: US 5315249 A

TITLE: Stabilized fast spin echo NMR pulse sequence

DATE-ISSUED: May 24, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Le Roux; Patrick L.	Gif/Yvette			FR
Hinks; Richard S.	Waukesha	WI		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 8. Document ID: US 4684891 A

L34: Entry 8 of 8 File: USPT Aug 4, 1987

US-PAT-NO: 4684891
DOCUMENT-IDENTIFIER: US 4684891 A

TITLE: Rapid magnetic resonance imaging using multiple phase encoded spin echoes in each of plural measurement cycles

DATE-ISSUED: August 4, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Feinberg; David A.	Berkeley	CA		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KMC | Draw D

Clear | Generate Collection | Print | Fwd Refs | Bkwd Refs | Generate OACS

Term	Documents
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CORRECTA	145
CORRECTAB	10
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CORRECTABJE	1
CORRECTABL	4
CORRECTABLE	9247
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CORRECTABLY	391
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Search Results - Record(s) 1 through 3 of 3 returned.

☐ 1. Document ID: US 5578923 A

Using default format because multiple data bases are involved.

L36: Entry 1 of 3 File: USPT Nov 26, 1996

US-PAT-NO: 5578923
DOCUMENT-IDENTIFIER: US 5578923 A

TITLE: T2 restoration and noise suppression of hybrid MR images using wiener and linear prediction techniques

DATE-ISSUED: November 26, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/309; 128/925

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 2. Document ID: US 5517122 A

L36: Entry 2 of 3 File: USPT May 14, 1996

US-PAT-NO: 5517122
DOCUMENT-IDENTIFIER: US 5517122 A

TITLE: T2 restoration and noise suppression of hybrid MR images using Wiener and linear prediction techniques

DATE-ISSUED: May 14, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Chen; Haiguang	San Francisco	CA		

US-CL-CURRENT: 324/322; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

☐ 3. Document ID: US 4684891 A

L36: Entry 3 of 3 File: USPT Aug 4, 1987

US-PAT-NO: 4684891
DOCUMENT-IDENTIFIER: US 4684891 A

TITLE: Rapid magnetic resonance imaging using multiple phase encoded spin echoes in each of plural measurement cycles

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INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Feinberg; David A.	Berkeley	CA		

US-CL-CURRENT: 324/309; 324/307

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

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ECHOES	20880
ECHOS	2888
ECHOE	140
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SIGNALS	2176866
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CORRECTA	145
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There are more results than shown above. Click here to view the entire set.

Display Format: - Change Format

L37: Entry 1 of 2 File: USPT Nov 26, 1996

DOCUMENT-IDENTIFIER: US 5578923 A
TITLE: T2 restoration and noise suppression of hybrid MR images using wiener and linear prediction techniques

Abstract Text (1):
Hybrid imaging (HI) sequences used for magnetic resonance (MR) imaging and inherently degraded by T2 effects and additive measurement noise are enhanced. Wiener filter and linear prediction (LP) technique is used to process HI MR signals in the spatial frequency domain (K-space) and the hybrid domain respectively. Based on the average amplitude symmetry constraint of the spin echo signal, the amplitude frequency response function of the T2 distortion is estimated and used in the Weiner filter for a global T2 amplitude restoration. Then a linear prediction technique is utilized to obtain local signal amplitude and phase estimates around discontinuities of the frequency response function of the equivalent T2 distortion filter. These estimates are used to make local amplitude and phase corrections. The effectiveness of this combined technique in correction T2 distortion and reducing the measurement noise is analyzed and demonstrated using experiments on both phantoms and humans.

Brief Summary Text (3):
This invention relates generally to the field of magnetic resonance (MR) imaging (MRI) utilizing NMR phenomena. It is particularly related to enhancement of MR imaging data acquired using hybrid imaging (HI) MR data acquisition sequences which include T2 and additive noise degradation effects.

Brief Summary Text (6):
In 1978, Mansfield et. al. demonstrated echo planar imaging (EPI) [P. Mansfield and P. G. Morris, "NMR Imaging in Biomedicine," in Advances in Magnetic Resonance, Edited by J. S. Waugh, Academic Press, New York, 1982]. The basic concept behind EPI is that successive spin echoes can be used to encode position information using just a single shot (i.e., a single NMR RF excitation data acquisition sequence). Because of high requirements on gradient coils and power supplies for achieving rapid echo train generation and some other problems, various hybrid imaging (HI) approaches, incorporating aspects of both conventional two-dimensional (2-D) FT imaging and EPI, have been proposed. [Hennig et al. J. Hennig, A. Nauerth and H. Friedberg, "RARE Imaging: A fast imaging Method for Clinical MR." Magne, Reson, Med., Vol. 3, pp. 823-33, 1986; Van Uijen et al. C. M. J. Van Uijen, J. H. Den Boef and F. J. J. Verschuren; Haacke et al. E. M. Haacke, F. H. Bearden, J. R. Clayton and N. R. Lingar, "Reduction of MR Imaging Time and Hybrid Fast Scan Technique," Radiology, Vol. 158, pp. 521-29, 1986; and others] These techniques use multiple (M) excitations and after each excitation, multiple (N) echoes are used to encode positional information. HI techniques are far less demanding on hardware and thus can be used to decrease imaging time without the cost and technical constraints of EPI.

Brief Summary Text (7):
Since in EPI and HI, phase encoding measurements acquired at different echo times are used to form an image, there are inherently T2 distortions in the acquired data along the phase encoding direction. Depending upon the phase encoding schemes used and the object under the study, loss of spatial resolution and/or contrast may be

introduced. [R. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI." Magne, Reson. Med., Vol. 28, pp. 9-24, 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "analysis of Hybird Imaging Techniques," Magne, Reson. Med., Vol. 26, pp. 155-73, 1992] Furthermore, there are discontinuities in the frequency response of the effective T2 distortion filter and these discontinuities generate ringing artifacts in the image. Techniques such as inverse filtering have been tried to reduce those T2 effects, based on some prior knowledge about the T2 values of the objects under study. The success of this approach is often limited by lack of knowledge about the T2 values and the existence of measurement noise. The problem of ringing artifacts caused by local discontinuities in the frequency response function of the T2 filter have not yet been successfully addressed.

Brief Summary Text (11):
It has now been discovered that a combined use of a Wiener filter and linear prediction (LP) to process HI images better moderates T2 and noise effects. In the first stage, based on the average amplitude symmetry constraint, a global T2 value of the object is estimated from acquired data and thus the amplitude frequency response function of the effective T2 distortion filter is determined. The Weiner filter is then used to make global T2 amplitude restoration and noise suppression in K-space. In the second stage, linear prediction is utilized to obtain local signal amplitude and phase estimates. That is, Weiner filter processed K-space signals are Fourier-transformed in the read-out direction to obtain a hybrid domain signal and LP is used to provide estimates of local signal amplitude and phase. These estimates are used to make local amplitude and phase corrections in the hybrid domain and thus reduce the effects caused by discontinuities of the T2 distortion filter frequency response. As a result of this two-stage processing, T2 effects on the image data can be reduced and, at the same time, measurement noise can also be suppressed.

Drawing Description Text (8):
FIGS. 7A-7D depict one dimensional images before and after T2 corrections;

Drawing Description Text (9):
FIGS. 8A and 8B show two-dimensional phantom images before and after T2 corrections; and

Drawing Description Text (10):
FIGS. 9A and 9B show two-dimensional human head images before and after T2 corrections.

Detailed Description Text (2):
FIG. 1 depicts a typical conventional MRI system that has been adapted so as to practice an exemplary embodiment of this invention. One example of such system is the Toshiba ACCESS.TM. MRI system. For example, it may comprise a rather large NMR polarizing magnet structure 10 which generates a substantially uniform homogeneous NMR polarizing magnetic field B.sub.0 within a patient imaging volume 12. A suitable carriage 14 is used for inserting the desired portion of patient 16 anatomy within the image volume 12. Magnetic NMR gradients in B.sub.0 can be selectively created by electromagnet gradient coils, NMR RF mutation pulses can be transmitted into the patient tissue within the image volume and NMR RF responses can be received from the patient tissue via suitable RF coil structures as will be appreciated by those in the art. A particular MRI data acquisition sequence of such magnetic gradient pulses, RF nutation pulses and NMR RF responses is conventionally achieved by an MRI sequence controller 18 controlling the usual array of gradient drivers 20. RF transmitter circuits 22 and RF receiver circuits 24, all suitable interfaced with electromagnetic and RF coils within the MRI system gantry. The received NMR RF responses are digitized and passed to an MRI image processor 26 which typically includes an array processor 28 and suitable computer program storage media 30 (e.g., RAM in silicon or magnetic media) wherein programs are

stored and selectively utilized so as to control the processing of acquired MR image data to produce digitized image displays on the CRT terminal 32. The control terminal 32 may also include suitable keyboard switches and the like for exerting operator control over the MRI sequence controller 18 and the interconnected cooperating MR image processor 26.

Detailed Description Text (4):
T2 effects on the quality of conventional MR images have been analyzed in many papers. The effects of T2 amplitude distortions on the HI images have also been investigated and reported in R. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI" Magne. Reson. Med., Vol. 28, pp. 9-24, 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne. Reson. Med., Vol. 26, pp. 155-73, 1992. The frequency responses of the effective T2 distortion filters and their effects on HI image quality continue to be of concern. Ringing artifacts introduced by discontinuities in the T2 filter frequency response are also of concern.

Detailed Description Text (5):
It is well known that MR images depend on multiple tissue parameters: the hydrogen density N(H), the longitudinal and transverse relaxation times, T1 and T2, and the pulse sequence parameters: repetition time TR and echo time TE. [D. A. Ortendahl and N. M. Hylton, "MRI Parameter Selection Techniques," in Magnetic Resonance Imaging, edited by C. L. Partian et. al., W. B. Saunders Company, 1988]. For example given a certain shaped object of uniform N(H), T1 and T2, the K-space MR signal S(k.sub.x,k.sub.y) acquired in a conventional spin-echo experiment can be written as ##EQU1## where s(n.sub.x,n.sub.y) is the observable magnetization from the object and (k.sub.x,k.sub.y) are spatial frequencies. Using the tissue parameters and the sequence parameters, equation (1) can also be expressed as

Detailed Description Text (6):
where the signal S(k.sub.x,k.sub.y) is determined only by the position, size and shape of the object. From equation (2), it can be seen that if TR and TE are fixed constants for different phase encode echoes (different k.sub.y values), there is no image distortion except for a fixed attenuation. In fact, along the frequency encoding (k.sub.x) direction, there is also a T2 distortion factor of e.sup.-Ts (k.sbsp.x.sup.)/T2 such that S(k.sub.x,k.sub.y) should be written as

Detailed Description Text (7):
with TS(k.sub.x) being the corresponding time for the sampling position k.sub.x. In practice, the durations of echoes are usually short compared to T2 times, so this effect is not visually apparent. [D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne, Reson. Med., Vol. 26, pp. 155-73, 1992]

Detailed Description Text (8):
When hybrid MR imaging techniques are used, different echoes in the echo train have different echo times TE(k.sub.y). Then, the acquired signal becomes

Detailed Description Text (10):
where S(k.sub.x,k.sub.y) is the K-space signal without T2 distortion

Detailed Description Text (12):
Thus, the effective T2 distortion filter equation (7) is separable in k.sub.x and k.sub.y directions. [D. E. Dudgeon and R. M. Mersereau, Multidimensional Digital Signal Processing, Prentice hall, Englewood Cliffs, N.J. 07632, 1981]. That is, it can be written as

Detailed Description Text (14):
This fact is useful for simplifying implementation of the T2 correction filter in practice. [D. E. Dudgeon and R. M. Mersereau, Multidimensional Digital Signal

Processing, Prentice Hall, Englewood Cliffs, N.J. 07732, 1981]. In MR imaging systems, noise results from multiple sources but it essentially consists of two major components: noise from the receiver circuits and noise from the excited tissues. These two components are affected by the system resonance frequency but are independent of echo times. Therefore, the T2 filter equation (7) has an effect on the signal but not on noise. [R. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI," Magne, Reson. Med., Vol. 28, pp. 9-24, 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne., Reson. Med., Vol. 26, pp. 155-73, 1992].

Detailed Description Text (15):
In the above formula for the T2 filter, absolute times TS(k.sub.x) and TE(k.sub.y) are used for the data sample S(k.sub.x,k.sub.y). For the purpose of explanation in the following discussion, relative times will be used for equation (7). In addition, it will be assumed that TS(k.sub.x)=0 for the first sample in the frequency encoding direction and TE(k.sub.y)=0 for the first echo in the echo train. This will normalize the T2 filter such that H(k.sub.x,k.sub.y).sub.max =1.0 but will not change the shapes of the frequency responses of the T2 distortion filter and therefore will not affect the structure of the resulting T2 correction filter.

Detailed Description Text (16):
Depending on the phase encoding schemes chosen, the T2 filter can have different frequency responses and thus have different effects on the resulting HI images. Consider, for instance, an M excitation HI sequence, each M containing N phase-encoded echoes to form an MN-line acquisition. If the earliest echo is assigned to the lowest spatial frequencies, with later echoes assigned to progressively higher spatial frequencies, the T2 filter has a low-pass frequency response in the k.sub.y direction. In this case, the image spatial resolution will be reduced. This case is similar to the blurring problems in many other imaging systems. On the other hand, if the earliest echo is assigned to the highest spatial frequencies, with later echoes assigned to progressively lower spatial frequencies, T2 filter has a high-pass frequency response in k.sub.y direction. Then, some edge enhancement may occur but the image contrast of large areas will be attenuated. In FIGS. 2 and 3 are shown the frequency responses H.sub.y (k.sub.y) of low-pass T2 filter and a high-pass T2 filter, respectively, for two HI sequences with M=64 and N=4. the object is assumed to have a T2 value of 70(ms). The echo times TE(i)=20.i ms are used for the echo number i=1,2,3,4, respectively, in the echo train. In these figures and the following discussions, the position of k.sub.y =129 corresponds to the zero-phase encode projection.

Detailed Description Text (17):
The T2 filter can also have frequency responses of band-pass, ramp and other shapes by using different sequence specifications. [R. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI," Magne, Reson. Med., Vol. 28, pp. 9-24 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne., Reson. Med., Vol. 26, pp. 155-73, 1992]. In FIG 4 is shown the frequency response of a band-pass T2 filter where the projections from the first echo are located in the middle frequency band. This encoding scheme has a better trade-off for image resolution and contrast that the low-pass and the high-pass encoding schemes. However, it usually generates more ringing artifacts because of the existence of bigger jumps in the frequency response shown in FIG. 4 than those shown in FIGS. 2 and 3. Note that for different phase encoding schemes, the T2 filter H.sub.x (k.sub.x) in the k.sub.x direction always has an exponential ramp frequency response.

Detailed Description Text (18):
In the above, only the amplitude frequency responses of T2 distortion filters is considered. It is known that for image representation, the phase components of an

image Fourier transform often have a more important role than the amplitude components. [M. H. Hayes, "The Reconstruction of a Multidimensional Sequence from the Phase or Magnitude of Its Fourier Transform," IEEE Trans. ASSP., Vol. 30, No. 2, pp. 140-54, 1982]. In MR imaging, in addition to amplitude distortions introduced by T2 effects, due to the imperfection of the practical imaging system and the difficulty of exact phase control in the HI sequence, there are also phase distortions caused by different phase shifts to the signal components from different echoes. These phase distortions will also generate ringing artifacts (even if there were no T2 amplitude distortions). Furthermore, the amplitude and phase discontinuities introduced by the T2 effects are signal dependent and thus cannot be smoothed by simple windowing. In FIG. 5A, the original image of two rectangular objects is shown, the image is shown with amplitude distortion only in FIG. 5B, the image with phase distortion only is shown in FIG. 5C, and the image with both amplitude and phase distortions is shown in FIG. 5D. The ampitude distortion is caused by an effective T2 distortion filter with a frequency response similar to that shown in FIG. 4. The phase distortion is caused by 90.degree. phase shift to the signal components from the second echo. The periods of these ringing artifacts are determined by the positions of the discontinuities in the T2 distortion filter frequency response. For the example of FIGS. 5A-5D, there is a big discontinuity in k.sub.y at position 96 which corresponds to the digital frequency f.apprxeq.0.13. Therefore, the period of the major ringing artifacts is about 7.7 pixels in this image of 256 pixels. These ringing artifacts seriously disturb accurate diagnosis using HI MR images. To improve the HI image quality, both the amplitude and phase distortions should be reduced.

Detailed Description Text (19):
In the previous discussion, one object with uniform tissue parameters has been assumed. In practice, an object under study usually has complicated distributions of these tissue parameters and the T2 values vary in different regions of the object. Signals from these different tissues are first distorted by the corresponding T2 distortion filters and then added together, further contaminated by measurement noise. Therefore, complete T2 correction is a challenging task. Nevertheless, as explained below, the use of a Wiener filter to make a global T2 amplitude correction (based on an estimated frequency response H(k.sub.x,k.sub.y) from the acquired data) provides a major improvement.

Detailed Description Text (20):
As previously shown, the T2 effects on images can be modeled as an original image distorted by a T2 distortion filter. If the frequency response of the T2 distortion filter is known, inverse filtering can recover the original image from the acquired data. That is,

Detailed Description Text (22):
The main problem with the inverse filter is its sensitivity of measurement noise. When the acquired data S(k.sub.x,k.sub.y) is in the form of the T2 distorted signal plus noise, which is the case in any practical imaging system, we have

Detailed Description Text (23):
where N(k.sub.x,k.sub.y) is the measurement noise component. Then the inverse filter will give ##EQU3## Since the noise component N(k.sub.x,k.sub.y) is not affected by the T2 filter and H(k.sub.x,k.sub.y)<1, we have N(k.sub.x,k.sub.y)>N (k.sub.x,k.sub.y). The measurement noise will be amplified in this T2 correction process and the resulting images are often not acceptable.

Detailed Description Text (24):
The Wiener algorithm provides a better solution to the T2 correction problem in a noisy environment. Let the signal s(n.sub.x,n.sub.y) and the noise n (n.sub.x,n.sub.y) be arbitrary, zero mean, random sequences, respectively. If the acquired sequence s(n.sub.x,n.sub.y) is modeled as

Detailed Description Text (25):
where h(n.sub.x,n.sub.y) is the distortion filter and II denotes 2-D convolution, then the best linear estimate s(n.sub.x,n.sub.y) for the original signal s (n.sub.x,n.sub.y) can be obtained from the distorted data s(n.sub.x,n.sub.y) using a Wiener filter w(n.sub.x, n.sub.y) in the sense that the mean square error between the original signal s(n.sub.x,n.sub.y) and the estimated signal s(n.sub.x,n.sub.y) defined by

Detailed Description Text (26):
is minimized where .epsilon. represents the expectation operation. The Wiener restoration filter w(n.sub.x,n.sub.y) has a frequency response W(k.sub.x,k.sub.y) in the form ##EQU4## where H(k.sub:x,k.sub.y) is the frequency response of the distortion filter h(x,y) and the superscript * represents conjugation. P.sub.ss (k.sub.x,k.sub.y) and P.sub.nn (k.sub.x, k.sub.y) are the power spectrums of the signal process s(x,y) and the noise process n(x,y), respectively. [R. C. Gonzalez and P. Wintz, Digital Image Processing, Addison-Wesley Publishing Company, 1988; A. K. Jain, Fundamentals of Digital Image Processing, Prentice Hall, Englewood Cliffs, N.J. 07632, 1989]

Detailed Description Text (28):
In MR imaging, the power spectrums of the signal and of the noise are not known and thus have to be estimated from the acquired data. Since S(k.sub.x,k.sub.y) is the inverse Fourier transform of the object, it usually has smaller values at higher spatial frequencies and thus the noise is often dominant in these frequencies. This will be especially true for hybrid fast MR imaging using a low-pass scheme since longer echo times further attenuate these high frequency signal components. Therefore, the average noise power P.sub.nn can be rather accurately estimated from the data S(k.sub.x,k.sub.y) on a support S.sub.n in the high spatial frequencies from the latest echo in the echo train using the standard formula ##EQU7## where N.sub.P is the total data number used for the P.sub.nn estimation on the data support S.sub.n. Since the measurement noise N(k.sub.x,k.sub.y) is often assumed to be white, we have the same noise power P.sub.nn everywhere in K-space. In practice, some DC component may exist. In this case, any such DC should be removed before making the estimation by equation (19).

Detailed Description Text (29):
Then the periodogram spectral estimation is used to obtain an estimate of the spectrum of the image data. [S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988] That is, we use the local data power P.sub.dd (k.sub.x,k.sub.y) defined by

Detailed Description Text (30):
to replace the signal power spectrum P.sub.ss (k.sub.x,k.sub.y) in equation (16) and thus obtain the frequency response of K-space Wiener T2 correction filter is ##EQU8## Since the frequency response H(k.sub.x,k.sub.y) of the T2 distortion filter is real in our discussion, the conjugate operation (*) and the absolute value operation (.vertline..multidot..vertline.) have been dropped in equation (21). The signal power can also be estimated by

Detailed Description Text (31):
This can be used in equation (21) instead of P.sub.dd (k.sub.x,k.sub.y) and will, in general, give stronger noise suppression. Once the frequency response of the Wiener filter is determined, the restored K-space image data can be obtained as ##EQU9##

Detailed Description Text (32):
One approach to determine the frequency response function H(k.sub.x,k.sub.y) from the acquired data is described below and from this the Wiener filter equation (21) can be complete specification for the image data restoration.

Detailed Description Text (33):
Since the frequency response H(k.sub.x,k.sub.y) of the T2 distortion filter depends on the spatial distribution and spin density of T2 values within the object under study, one cannot determine the exact frequency response of the T2 filter with just one data acquisition. Nevertheless, one approach for estimating H(k.sub.x,k.sub.y) from the acquired data based on the average amplitude symmetry constraint of the spin echo signal is set forth below.

Detailed Description Text (34):
It is well known from MR spectroscopy that the time-domain echo signal is symmetric around its peak if the T2 value is much longer than the echo observation time. In MR imaging, when the same TE is used for all phase encode (k.sub.y) values, ideally, the K-space data S(129+k.sub.x,129+k.sub.y) and S(129-k.sub.x,129-k.sub.y) should be conjugate symmetric about the peak signal position (k.sub.x =129,k.sub.y= 129) in a 256.times.256 data matrix and have the same amplitude values if the T2 values of the object are much longer than the sampling window width in the frequency encoding (read-out, k.sub.x) direction. [D. A. Feinberg, J. D. Hale, J. C. Watts, L. Kaufman and A. Mark, "Halving MR Imaging Time by Conjugation: Demonstration at 3.5 kG." Radiology 164, pp. 527-31, 1986] When the T2 distortion effect exists, data amplitude distributions are biased. Therefore, one can estimate a global T2 value from the acquired data using this symmetric constraint.

Detailed Description Text (35):
In FIG. 6 is shown a profile of the K-space data .vertline.S(k.sub.x,129).vertline. for a rectangular object. It is symmetric about and peaked at k.sub.x =129. The curve on the top part of FIG. 6 is the exponential function e.sup.-t/T2 with T2=100 ms and the sampling window width T.sub.x =10 ms. If there is no noise and there is only a global T2 effect, any two values of .vertline.S(k.sub.x,129).vertline. from the two points symmetric about the peak would provide an estimate of the global T2 value. But there are many factors, including measurement noise and phase shifts, which affect the amplitude symmetry property. Therefore, one may first calculate the sums of the acquired signal amplitude .vertline.S(k.sub.x,129).vertline. over two equal time spans symmetric about the echo peak. For example, if S(k.sub.x,129) has N.sub.x samples in the k.sub.x direction with the sampling window time T.sub.x, we calculate the average amplitude A1 and A2 over a fraction of T.sub.x as ##EQU10## where N.sub.T2x is the number of data samples in the T2 estimate window T.sub.w. An average T2 value can be calculated from these two amplitude values as

Detailed Description Text (36):
where T.sub.p is the period of the data sampling in the k.sub.x direction and thus T.sub.x=N.sub.x T.sub.p. From equation (1), it is known that ##EQU11## and thus the echo signal S(k.sub.x,129) is composed of contributions from all elements s (n.sub.x,n.sub.y) of the object. Therefore, (25) provides only a global T2 estimate of the object.

Detailed Description Text (37):
The derivation of equation (25) is based on an assumption that the signal has a flat Fourier spectrum over the T2 estimate window T.sub.w. For real signal spectrums with arbitrary shapes, the estimated global T2 will deviate from the true global T2 value and the difference between the true T2 value and estimated T2 value is affected by the width T.sub.w and the shape of the signal spectrum S (k.sub.x,k.sub.y). When there is no noise and no phase error, a shorter estimate window generates a smaller estimate difference (bias) from the true T2 value. When there are noise and other disturbing factors, the estimate window cannot be too short. It is the summation (low-pass filtering) process that reduces the disturbing effects such as noise and phase shifts, and helps obtain a stable T2 estimation. This is a typical tradeoff between the bias and the variance in the estimation problem [L. L. Scharf, Statistical Signal Processing: Detection, Estimation, and Time Series Analysis, Addison-Wesley Publishing Company, 1990] In practice, the T2 estimate window size T.sub.w should be adjusted according to the MR imaging

conditions.

Detailed Description Text (40):
A1 and A2 could also be calculated over a wider window in k.sub.y direction such that ##EQU13## where N.sub.T2,y determines the width (2N.sub.T2,y +1) of the T2 estimate window in the k.sub.y direction. When N.sub.T2,y =0, equation (29) reduces to equation (27). After a global T2 value has been estimated, the frequency response H(k.sub.x,k.sub.y) of the T2 distortion filter can be determined using equation (7), according to the MR sequence specification of echo times TE(k.sub.y) and the data sampling period T.sub.p.

Detailed Description Text (41):
The approach discussed above makes the estimation of H(k.sub.x,k.sub.y) from the acquired image data directly and thus does not require extra data acquisition. But this approach can only estimate the amplitude distortion function of the T2 filter and therefore the resulting Weiner filter based on these estimates can only reduce the amplitude distortion caused by the T2 effect.

Detailed Description Text (42):
Linear prediction techniques have been successfully used for time-series analysis, high resolution spectrum estimation, speech and image signal coding, and many other applications [S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988] The technique has also been used for MR data extrapolation to reduce the truncation artifacts and improve spatial resolution [M. R. Smith, S. T. Nichols, R. M. Henkelman and M. L. Wood, "Application of Autoregressive Moving Average Parametric Modeling in Magnetic Resonance Image Reconstruction," IEEE Trans Med., Imag., Vol. 5, pp. 132-39, 1986; J. F. Martin and C. F. Tirendi, "Modified Linear Prediction Modeling in Magnetic Resonance Imaging," J. Magn. Reson, Vol. 82, pp. 392-99, 1989; E. M. Haacke, Z. Liang and S. H. Izen, "Super Resolution Reconstruction Through Object Modeling and Parameter Estimation," IEEE Trans. ASSP., Vol. 37, pp. 392-95, 1989]. the application of linear prediction to local T2 correction of both amplitude and phase distortions is discussed below.

Detailed Description Text (45):
Assume that the global T2 corrected MR data by the Wiener filter are represented by a 2-D function S(k.sub.x,k.sub.y) where k.sub.x is the index of sampling points in the frequency encode direction and k.sub.y is the index of sampling points in the phase encode direction. First, the inverse Fourier transform of the time-domain data S(k.sub.x, k.sub.y) is taken with respect to k.sub.x to obtain the hybrid-domain data set X(n.sub.x,k.sub.y) and then the data X(n.sub.x,k.sub.y) is considered for each n.sub.x value. Given the discontinuity positions in the frequency response of T2 distortion filters, the linear prediction is used to produce data samples across these positions. For example, if the T2 distortion filter is as shown in FIG. 2, a total of 31 data samples from X(n.sub.x, 130) to X (n.sub.x, 160) can be used to predict data samples on the other side of the discontinuity position by forward linear prediction, starting from k.sub.y =161. The prediction is performed from data samples of lower frequency to data samples of higher frequency since low frequency data samples usually have higher signal-to-noise ratio and therefore the resulting predicted data have lower prediction errors.

Detailed Description Text (47):
This local phase distortion factor is then used to make a local zero-order phase correction by multiplying all the old data samples of the second echo from k.sub.y =161 to k.sub.y =192. The resulting phase corrected data are denoted as X (n.sub.x,k.sub.y). That is

Detailed Description Text (48):
The estimated local phase distortion factor .psi.(n.sub.x) from different column

data X(n.sub.x,k.sub.y) can be first low-pass filtered and the resulting smoothed phase estimates are then used in equation (34) for phase correction. This enhances the stability of local phase correction.

Detailed Description Text (49):
Since the prediction usually has quite precise results for data samples close to the starting position (k.sub.y =161) in the echo train boundary and becomes less accurate as k.sub.y moves away from the boundary, predicted data X(nx,k.sub.y) are combined with zero-order phase corrected data x(n.sub.x,k.sub.y) as a weighted average over a data merging band to obtain local amplitude and phase corrected data X(n.sub.x,k.sub.y). That is,

Detailed Description Text (50):
where N.sub.m is the width of the data merging band. Two weighting factors w.sub.1 (k.sub.y) and w.sub.2 (k.sub.y) are some monotonously decreasing and increasing functions of k.sub.y, respectively. They are related as w.sub.1 (k.sub.y)+w.sub.2 (k.sub.y)=1.0 for all k.sub.y. The use of these weighting functions are intended to ensure a smooth transition from the predicted data to the originally observed data. In this study, w.sub.1 (k.sub.y) and w.sub.2 (k.sub.y) are chosen as linear functions of k.sub.y and thus ##EQU16## Although other weighting functions such as quadratic and exponential functions have been tried to make different weightings on the two data sets, it has been found that they do not make a significant performance difference. The determination of the data merging band N.sub.m is affected by several considerations. A wider merging band usually provides stronger ringing artifact reduction but it also results in bigger changes in the original signal spectrum. Also, its value is limited by the data number in each echo group. A band around ten data samples gives good results for four-echo HI images of 256 phase encoding projections.

Detailed Description Text (51):
A similar local amplitude and phase correction is performed around other discontinuity positions in the data set X(n.sub.x,k.sub.y) except that for the discontinuity positions of k.sub.y <129, a backward linear prediction is used to estimate data samples across these positions (based also on the data samples with lower frequencies). Finally, the inverse Fourier transform of the hybrid domain data X(n.sub.x,k.sub.y) is taken with respect to k.sub.y to reconstruct the HI image. As shown by the subsequently discussed examples, this processing effectively reduces effects from amplitude and phase discontinuities in the frequency response of the effective T2 distortion filter and therefore reduces ringing artifacts in the reconstructed HI image.

Detailed Description Text (53):
Because the phase error has not been corrected, it is noted that image resolution loss and ringing artifacts still exist. If local amplitude and phase correction is applied to the distorted image data directly, a local corrected image is obtained as shown in FIG. 7C with improved resolution and reduced ringing artifacts. In this case, 30 data samples were used to calculate the autocorrelation function (ACF) with a biased ACF estimator [L. B. Jackson, Digital Filters and Signal Processing, Second Edition, Kluwer Academic Publishers, 1989] For the predictor order selection, there are many techniques available [S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988] The model order selection rule N/3<p<N/2 by Ulrych and Bishop [T. J. Ulrych and T. N. Bishop, "Maximum Entropy Spectral Analysis and Autoregressive Decomposition," Rev. Geophys. Space Phys., Vol. 13, pp. 183-200, 1975] where N is the number of data samples used for the ACF estimation was used here. Therefore, the order p of the linear predictor for this case has been chosen to be 12. The predictor coefficients are determined with the Levinson-Durbin algorithm. The new data samples can be predicted using equations (30) or (31) based on known data samples and previously predicted data samples. The estimated local phase distortion using equation (33) from the forward predicted data X(161) and the old data sample

X(161) is 97.15.degree. and the estimated local phase distortion is 92.54.degree. from the backward predicted data X(96) and the old data X(96). The phase estimate error is less than 10%. As described above, these phase estimates are used to make the zero-order phase correction on data of k.sub.y >161 and k.sub.y <96, respectively. Then the phase corrected data are merged with the estimated data to form a new data set for final image reconstruction. A data merging band of ten data samples has been used for all discontinuity positions in FIG. 4.

Detailed Description Text (54):
Finally, the local amplitude and phase correction technique is employed on data processed by the global T2 amplitude correction. The resulting corrected HI image by this two-stage procedure is given in FIG. 7D. It is clear from these figures that the original HI image degradation caused by amplitude attenuation, resolution loss and ringing artifacts has been effectively improved by this combined global and local data processing technique (except near the sharp edges of the objects).

Detailed Description Text (55):
The proposed method has been applied to an image of a phantom acquired in a low-field-strength (0.064 T) permanent magnet imager (ACCESS.TM. Toshiba-America MRI Inc.). The phantom object is a rectangular container full of mineral oil with a T2 value of about 100 ms.

Detailed Description Text (56):
The asymmetric Fourier imaging (AFI) approach has been employed to further reduce the imaging time. A total of 144 phase encode projections has been acquired by 36 data acquisition shots using a four-echo band-pass sequence with echo times TE(i) =10.i ms for i=1,2,3,4. The data is then conjugated to a full image size of 256.times.256 pixels. The resulting T2 filter has a frequency response similar to that shown in FIG. 4 and phase encode is in the vertical direction. Because of amplitude and phase distortions introduced by the T2 distortion filter, the reconstructed image has serious ringing artifacts along the phase encoding direction as evidenced in FIG. 8A. For demonstration purposes, one of the worst cases of ringing artifacts is illustrated.

Detailed Description Text (57):
With a global T2 estimate window of N.sub.T2,x =64,N.sub.T2y =2 and the root factor .alpha.=0.5, the global T2 value estimated using equations (27) and (28) is 127 ms. This estimated T2 value is used in a Wiener filter for global amplitude correction on the originally acquired 144 projections. Now, 21 data samples are used for the ACF estimate and the predictor order is ten. For the AFI data case, both the global and the local corrections should be performed before conjugation and this will save computations needed for the data correction. For the given AFI data of 144 phase projections, the saving in the global T2 amplitude correction is about 44% compared with the full size data of 256 phase projections. For local corrections, the saving is 50% since only three instead of six discontinuities need to be processed. T2 corrected data are conjugated to the full image size and the improved image as shown in FIG. 8B is obtained which has much less ringing artifact. The standard deviation of the noise has been reduced form 144 to 55 by the Wiener filtering.

Detailed Description Text (58):
As a further example, one image from a sequence with both sin echo and gradient echo has been tested. The sequence consists of three echoes in the echo train, one spin echo and two gradient echoes. The spin echo has an echo time 25 ms and generates the signal components S(k.sub.x,k.sub.y) for 86<k.sub.y <170. The first gradient echo has an echo time 19 ms and generates signal components S (k.sub.x,k.sub.y) for 1<k.sub.y <85. the second gradient echo has an echo time 32 ms, generating the remaining signal components. The image was acquired in a high-field-strength (1.5 T) MR image (Toshiba MRT200/FXIII). Similar to the fast spin echo case, because of T2 and T2* effects, the ringing artifacts in the

reconstructed image are obvious as shown in FIG. 9A, which seriously degrade the image quality. In this image, phase encoding is in the horizontal direction.

Detailed Description Text (59):
Although the echo times given above seem to define a ramp T2 filter, the actual T2filter has a low-pass shape since the dephasing effects T2*effects) caused by field inhomogeneity results in faster decay for the gradient echo signal components than from the T2 effect. In this case, the T2 amplitude attenuation factors A.sub.g.spsb.1 for the first gradient echo and A.sub.g.spsb.2 for the second gradient echo are estimated using signals from three extra zero phase encoding projections in the end of the sequence. The are determined by ##EQU17## where S.sub.s (k.sub.x,129) is the zero-phase encoding signal from the standard spin echo, S.sub.g.spsb.1 (k.sub.x,129) and S.sub.g.spsb.2 (k.sub.x,129) are the zero-phase encoding signals from the fist and the second gradient echoes, respectively. For the given image, estimated attenuation factors are A.sub.g.spsb.1 =0.84 and A.sub.g.spsb.2 =0.72 with respect to the spin echo signal. These attenuation factors were used in the Wiener filter to make a global amplitude correction. This estimation method for global T2 amplitude attenuation factors can also be used for the FSE case. Then, the Wiener filter and the linear prediction are applied to this image and the resulting image after the two-stage processing is shown in FIG. 9B (which has improved resolution and fewer ringing artifacts). The standard deviation of noise in the host homogeneous tissue region has also been reduced from 149 to 118.

Detailed Description Text (60):
In the above experiments, the Wiener filter and the linear prediction technique have been used together to effectively improve HI image quality. For some applications, the Wiener filter can be used only to perform partial global T2 amplitude correction. For example, the sequence with a band-pass frequency response as shown in FIG. 4 is often used to emphasize signal components with middle range frequencies and obtain an edge enhanced image. In this case, for preserving the desired objective, a global Wiener T2 amplitude correction can be employed only for signal components from the third and fourth echoes to further increase high frequency components (leaving the low frequency components from the second echo untouched). Then the linear prediction technique is used to reduce ringing artifacts.

Detailed Description Text (61):
The combined use of a Wiener filter and a linear prediction technique for T2 restoration and noise suppression has been presented. The effectiveness of this method in improving HI image quality has been illustrated using some experimental results in which only estimated global T2 value (and therefore a simple frequency response function H(k.sub.x,k.sub.y) of T2 filter) is available for the T2 amplitude restoration. This simplified model may result in under and/or over compensations for certain objects in a given image. Methods for more accurate estimate of T2 values and distributions are required to obtain better T2 correction. For local amplitude and phase correction, the simple linear prediction has been used for computation simplification. Other modeling techniques such as autoregressive moving average (ARMA) [M. R. Smith, S. T. Nichols, R. M. Henkelman and M. L. Wood, "Application of Autoregressive Moving Average Parametric Modeling in Magentic Resonance Image Reconstruction," IEEE Trans. Med. Imag., Vol. 5, pp. 132-39, 1986; S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988; L. B. Jackson, Digital Filters and Signal Processing, Second Edition, Kluwer Academic Publishers, 1989] and nonlinear prediction with neural network [H. Yan and J. Mao, "Data Truncation Artifact Reduction in MR Imaging Using a Multilayer Neural Network," IEEE Trans. Med. Imag., Vol. 12, pp. 73-77, 1993] can be employed to improve the results.

Detailed Description Text (62):
As indicated above, this invention may be implemented by suitably programming the

image data processor of a conventional MRI system so as to implement the above-described two-stage T2 correction (global, then local) process. The program may be provided as optional additional subroutines if desired or incorporated as permanently used non-optional added data processing routines. In any such case, the programs may be conventionally written by those skilled in the art to strightforwardly implement the above-described processes. Accordingly, no further detailed description of such computer programs is believed to be necessary.

Detailed Description Paragraph Equation (7):
H(k.sub.x,k.sub.y)=e.sup.-(TS(k.sbsp.x.sup.)/T2 e.sup.-TE(k.sbsp.y.sup.)/T2 =H.sub.x (k.sub.x)H.sub.y (k.sub.y) (8)

Detailed Description Paragraph Equation (8):
H.sub.x (k.sub.x)=e.sup.-TS(k.sbsp.x.sup.)/T2, H.sub.y (k.sub.y)=e.sup.-TE (k.sbsp.y.sup.)/T2 (9)

Detailed Description Paragraph Equation (9):
S(k.sub.x,k.sub.y)=H.sup.I (k.sub.x, k.sub.y)S(k.sub.x, k.sub.y) (10)

Detailed Description Paragraph Equation (10):
S(k.sub.x, k.sub.y)=H(k.sub.x,k.sub.y)S(k.sub.x,k.sub.y)+N(k.sub.x,k.sub.y) (12)

Detailed Description Paragraph Equation (19):
X(n.sub.x,k.sub.y)=exp(+i.psi.(n.sub.x)).multidot.X(n.sub.x,k.sub.y) for 161<k.sub.y <192 (34)

Detailed Description Paragraph Equation (20):
X(n.sub.x,k.sub.y)=w.sub.1 (k.sub.y)X(n.sub.x,k.sub.y)+w.sub.2 (k.sub.y)X (n.sub.x,k.sub.y) for 161<k.sub.y <(161+N.sub.m) (35)

CLAIMS:

1. A method for enhancing MRI data acquired at different NMR echo times to compensate for T2 and noise distortions, said method comprising the steps of:

deriving linear predictions of estimated local amplitude and phase for one-dimensional Fourier-transformations of MRI data in the spatial frequency domain, and

utilizing said linear predictions to make local amplitude and phase corrections to such data during MR image reconstruction processing.

2. Apparatus for enhancing MRI data acquired at different NMR echo times to compensate for T2 and noise distortions, said apparatus comprising:

means for deriving linear predictions of estimated local amplitude and phase for one-dimensional Fourier-transformations of MRI data in the spatial frequency domain, and

means for utilizing said linear predictions to make local amplitude and phase corrections to such data during MR image reconstruction processing.

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L37: Entry 2 of 2 File: USPT May 14, 1996

DOCUMENT-IDENTIFIER: US 5517122 A
TITLE: T2 restoration and noise suppression of hybrid MR images using Wiener and linear prediction techniques

Abstract Text (1):
Hybrid imaging (HI) sequences used for magnetic resonance (MR) imaging and inherently degraded by T2 effects and additive measurement noise are enhanced. Wiener filter and linear prediction (LP) technique is used to process HI MR signals in the spatial frequency domain (K-space) and the hybrid domain respectively. Based on the average amplitude symmetry constraint of the spin echo signal, the amplitude frequency response function of the T2 distortion is estimated and used in the Wiener filter for a global T2 amplitude restoration. Then a linear prediction technique is utilized to obtain local signal amplitude and phase estimates around discontinuities of the frequency response function of the equivalent T2 distortion filter. These estimates are used to make local amplitude and phase corrections. The effectiveness of this combined technique in correcting T2 distortion and reducing the measurement noise is analyzed and demonstrated using experiments on both phantoms and humans.

Brief Summary Text (3):
This invention relates generally to the field of magnetic resonance (MR) imaging (MRI) utilizing NMR phenomena. It is particularly related to enhancement of MR imaging data acquired using hybrid imaging (HI) MR data acquisition sequences which include T2 and additive noise degradation effects.

Brief Summary Text (6):
In 1978, Mansfield et al. demonstrated echo planar imaging (EPI) [P. Mansfield and P. G. Morris, "NMR Imaging in Biomedicine," in Advances in Magnetic Resonance, Edited by J. S. Waugh, Academic Press, New York, 1982] The basic concept behind EPI is that successive spin echoes can be used to encode position information using just a single shot (i.e., a single NMR RF excitation data acquisition sequence). Because of high requirements on gradient coils and power supplies for achieving rapid echo train generation and some other problems, various hybrid imaging (HI) approaches, incorporating aspects of both conventional two dimensional (2-D) FT imaging and EPI, have been proposed. [Hennig et al. J. Hennig, A. Nauerth and H. Friedberg; "RARE Imaging: A Fast Imaging Method for Clinical MR." Magne, Reson, Med., Vol. 3, pp. 823-33, 1986; Van Uijen et al. C. M. J. Van Uijen, J. H. Den Boer and F. J. J. Verschuren; Haacke et al. E. M. Haacke, F. H. Bearden, J. R. Clayton and N. R. Lingar, "Reduction of MR Imaging Time and Hybrid Fast Scan Technique," Radiology, Vol. 158, pp. 521-29, 1986; and others] These techniques use multiple (M) excitations and after each excitation, multiple (N) echoes are used to encode positional information. HI techniques are far less demanding on hardware and thus can be used to decrease imaging time without the cost and technical constraints of EPI.

Brief Summary Text (7):
Since in EPI and HI, phase encoding measurements acquired at different echo times are used to form an image, there are inherently T2 distortions in the acquired data

along the phase encoding direction. Depending upon the phase encoding schemes used and the object under the study, loss of spatial resolution and/or contrast may be introduced. [R. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI." Magne, Reson. Med., Vol. 28, pp. 9-24, 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "analysis of Hybrid Imaging Techniques, Magne, Reson. Med., Vol. 26, pp. 155-73, 1992] Furthermore, there are discontinuities in the frequency response of the effective T2 distortion filter and these discontinuities generate ringing artifacts in the image. Techniques such as inverse filtering have been tried to reduce these T2 effects, based on some prior knowledge about the T2 values of the objects under study. The success of this approach is often limited by lack of knowledge about the T2 values and the existence of measurement noise. The problem of ringing artifacts caused by local discontinuities in the frequency response function of the T2 filter have not yet been successfully addressed.

Brief Summary Text (11):
It has now been discovered that a combined use of a Wiener filter and linear prediction (LP) to process HI images better moderates T2 and noise effects. In the first stage, based on the average amplitude symmetry constraint, a global T2 value of the object is estimated from acquired data and thus the amplitude frequency response function of the effective T2 distortion filter is determined. The Wiener filter is then used to make global T2 amplitude restoration and noise suppression in K-space. In the second stage, linear prediction is utilized to obtain local signal amplitude and phase estimates. That is, Wiener filter processed K-space signals are Fourier-transformed in the read-out direction to obtain a hybrid domain signal and LP is used to provide estimates of local signal amplitude and phase. These estimates are used to make local amplitude and phase corrections in the hybrid domain and thus reduce the effects caused by discontinuities of the T2 distortion filter frequency response. As a result of this two-stage processing, T2 effects on the image data can be reduced and, at the same time, measurement noise can also be suppressed.

Drawing Description Text (8):
FIGS. 7A-7D depict one dimensional images before and after T2 corrections;

Drawing Description Text (9):
FIGS. 8A and 8B show two-dimensional phantom images before and after T2 corrections; and

Drawing Description Text (10):
FIGS. 9A and 9B show two-dimensional human head images before and after T2 corrections.

Detailed Description Text (2):
FIG. 1 depicts a typical conventional MRI system that has been adapted so as to practice an exemplary embodiment of this invention. One example of such system is the Toshiba ACCESS.TM. MRI system. For example, it may comprise a rather large NMR polarizing magnet structure 10 which generates a substantially uniform homogeneous NMR polarizing magnetic field B.sub.o within a patient imaging volume 12. A suitable carriage 14 is used for inserting the desired portion of patient 16 anatomy within the image volume 12. Magnetic NMR gradients in B.sub.0 can be selectively created by electromagnet gradient coils, NMR RF mutation pulses can be transmitted into the patient tissue within the image volume and NMR RF responses can be received from the patient tissue via suitable RF coil structures as will be appreciated by those in the art. A particular MRI data acquisition sequence of such magnetic gradient pulses, RF nutation pulses and NMR RF responses is conventionally achieved by an MRI sequence controller 18 controlling the usual array of gradient drivers 20, RF transmitter circuits 22 and RF receiver circuits 24, all suitably interfaced with electromagnetic and RF coils within the MRI system gantry. The received NMR RF responses are digitized and passed to an MRI image processor 26

which typically includes an array processor 28 and suitable computer program storage media 30 (e.g., RAM in silicon or magnetic media) wherein programs are storm and selectively utilized so as to control the processing of acquired MR image dam to produce digitized image displays on the CRT terminal 32. The control terminal 32 may also include suitable keyboard switches and the like for exerting operator control over the MRI sequence controller 18 and the interconnected cooperating MR image processor 26.

Detailed Description Text (4):
T2 effects on the quality of conventional MR images have been analyzed in many papers. The effects of T2 amplitude distortions on the HI images have also been investigated and reported in R. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI," Magne. Reson. Med., Vol. 28, pp. 9-24, 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne. Reson. Med., Vol. 26, pp. 155-73, 1992. The frequency responses of the effective T2 distortion filters and their effects on HI image quality continue to be of concern. Ringing artifacts introduced by discontinuities in the T2 filter frequency response are also of concern.

Detailed Description Text (5):
It is well known that MR images depend on multiple tissue parameters: the hydrogen density N(H), the longitudinal and transverse relaxation times, T1 and T2, and the pulse sequence parameters: repetition time TR and echo time TE. [D. A. Ortendahl and N. M. Hylton, "MRI Parameter Selection Techniques," in Magnetic Resonance Imaging, edited by C. L. Partain et al., W. B. Saunders Company, 1988]. For example, given a certain shaped object of uniform N(H), T1 and T2, the K-space MR signal S(k.sub.x,k.sub.y) acquired in a conventional spin-echo experiment can be written as ##EQU1## where s(n.sub.x,n.sub.y) is the observable magnetization from the object and (k.sub.x,k.sub.y) are spatial frequencies. Using the tissue parameters and the sequence parameters, equation (1) can also be expressed as

Detailed Description Text (6):
where the signal S(k.sub.x,k.sub.y) is determined only by the position, size and shape of the object. From equation (2), it can be seen that if TR and TE are fixed constants for different phase encode echoes (different k.sub.y values), there is no image distortion except for a fixed attenuation. In fact, along the frequency encoding (k.sub.x) direction, there is also a T2 distortion factor of .sub.e -TS (k.sub.x)/T2 such that S(k.sub.x,k.sub.y) should be written as

Detailed Description Text (7):
with TS(k.sub.x) being the corresponding time for the sampling position k.sub.x. In practice, the durations of echoes are usually short compared to T2 times, so this effect is not visually apparent. [D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne, Reson. Med., Vol. 26, pp. 155-73, 1992]

Detailed Description Text (8):
When hybrid MR imaging techniques are used, different echoes in the echo train have different echo times TE(k.sub.y). Then, the acquired signal becomes

Detailed Description Text (10):
where S(k.sub.x,k.sub.y) is the K-space signal without T2 distortion

Detailed Description Text (12):
Thus, the effective T2 distortion filter equation (7) is separable in k.sub.x and k.sub.y directions. [D. E. Dudgeon and R. M. Mersereau, Multidimensional Digital Signal Processing, Prentice Hall, Englewood Cliffs, N.J. 07632, 1981]. That is, it can be written as

Detailed Description Text (14):

This fact is useful for simplifying implementation of the T2 correction filter in practice. [D. E. Dudgeon and R. M. Merserean, Multidimensional Digital Signal Processing, Prentice Hall, Englewood Cliffs, N.J. 07632, 1981]. In MR imaging systems, noise results from multiple sources but it essentially consists of two major components: noise from the receiver circuits and noise from the excited tissues. These two components are affected by the system resonance frequency but are independent of echo times. Therefore, the T2 filter equation (7) has an effect on the signal but not on noise. JR. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI," Magne, Reson. Med., Vol. 28, pp. 9-24, 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne., Reson. Med., Vol. 26, pp. 155-73, 1992].

Detailed Description Text (15):
In the above formula for the T2 filter, absolute times TS(k.sub.x) and TE(k.sub.y) are used for the data sample S(k.sub.x,k.sub.y). For the purpose of explanation in the following discussion, relative times will be used for equation (7). In addition, it will be assumed that TS(k.sub.x)=0 for the first sample in the frequency encoding direction and TE(k.sub.y)=0 for the first echo in the echo train. This will normalize the T2 filter such that H(k.sub.x,k.sub.y).sub.max =1.0 but will not change the shapes of the frequency responses of the T2 distortion filter and therefore will not affect the structure of the resulting T2 correction filter.

Detailed Description Text (16):
Depending on the phase encoding schemes chosen, the T2 filter can have different frequency responses and thus have different effects on the resulting HI images. Consider, for instance, an M excitation HI sequence, each M containing N phase-encoded echoes to form an MN-line acquisition. If the earliest echo is assigned to the lowest spatial frequencies, with later echoes assigned to progressively higher spatial frequencies, the T2 filter has a low-pass frequency response in the k.sub.y direction. In this case, the image spatial resolution will be reduced. This case is similar to the blurring problems in many other imaging systems. On the other hand, if the earliest echo is assigned to the highest spatial frequencies, with later echoes assigned to progressively lower spatial frequencies, T2 filter has a high-pass frequency response in k.sub.y direction. Then, some edge enhancement may occur but the image contrast of large areas will be attenuated. In FIGS. 2 and 3 are shown the frequency responses H.sub.y (k.sub.y) of a low-pass T2 filter and a high-pass T2 filter, respectively, for two HI sequences with M=64 and N=4. The object is assumed to have a T2 value of 70 (ms). The echo times TE(i)=20 .i ms are used for the echo number i=1,2,3,4, respectively, in the echo train. In these figures and the following discussions, the position of k.sub.y =129 corresponds to the zero-phase encode projection.

Detailed Description Text (17):
The T2 filter can also have frequency responses of band-pass, ramp and other shapes by using different sequence specifications. [R. T. Constable and J. C. Gore, "The Loss of Small Objects in Variable TE Imaging: Implications for FSE, RARE, and EPI," Magne,Reson. Med., Vol. 28, pp. 9-24, 1992; D. A. Ortendahl, L. Kaufman and D. M. Kramer, "Analysis of Hybrid Imaging Techniques," Magne.,Reson. Med., Vol. 26, pp. 155-73, 1992]. In FIG. 4 is shown the frequency response of a band-pass T2 filter where the projections from the first echo are located in the middle frequency band. This encoding scheme has a better trade-off for image resolution and contrast than the low-pass and the high-pass encoding schemes. However, it usually generates more ringing artifacts because of the existence of bigger jumps in the frequency response shown in FIG. 4 than those shown in FIGS. 2 and 3. Note that for different phase encoding schemes, the T2 filter H.sub.x (k.sub.x) in the k.sub.x direction always has an exponential ramp frequency response.

Detailed Description Text (18):

In the above, only the amplitude frequency responses of T2 distortion filters is considered. It is known that for image representation, the phase components of an image Fourier transform often have a more important role than the amplitude components. [M. H. Hayes, "The Reconstruction of a Multidimensional Sequence from the Phase or Magnitude of Its Fourier Transform," IEEE Trans. ASSP., Vol. 30, No. 2, pp. 140-54, 1982]. In MR imaging, in addition to amplitude distortions introduced by T2 effects, due to the imperfection of the practical imaging system and the difficulty of exact phase control in the HI sequence, there are also phase distortions caused by different phase shifts to the signal components from different echoes. These phase distortions will also generate ringing artifacts (even if there were no T2 amplitude distortions). Furthermore, the amplitude and phase discontinuities introduced by the T2 effects are signal dependent and thus cannot be smoothed by simple windowing. In FIG. 5A, the original image of two rectangular objects is shown, the image is shown with amplitude distortion only in FIG. 5B, the image with phase distortion only is shown in FIG. 5C, and the image with both amplitude and phase distortions is shown in FIG. 5D. The amplitude distortion is caused by an effective T2 distortion filter with a frequency response similar to that shown in FIG. 4. The phase distortion is caused by a 90.degree. phase shift to the signal components from the second echo. The periods of these tinging artifacts are determined by the positions of the discontinuities in the T2 distortion filter frequency response. For the example of FIGS. 5A-5D, there is a big discontinuity in k.sub.y at position 96 which corresponds to the digital frequency f.apprxeq.0.13. Therefore, the period of the major tinging artifacts is about 7.7 pixels in this image of 256 pixels. These tinging artifacts seriously disturb accurate diagnosis using HI MR images. To improve the HI image quality, both the amplitude and phase distortions should be reduced.

Detailed Description Text (19):
In the previous discussion, one object with uniform tissue parameters has been assumed. In practice, an object under study usually has complicated distributions of these tissue parameters and the T2 values vary in different regions of the object. Signals from these different tissues are first distorted by the corresponding T2 distortion filters and then added together, further contaminated by measurement noise. Therefore, complete T2 correction is a challenging task. Nevertheless, as explained below, the use of a Wiener filter to make a global T2 amplitude correction (based on an estimated frequency response H(k.sub.x,k.sub.y) from the acquired data) provides a major improvement.

Detailed Description Text (20):
As previously shown, the T2 effects on images can be modeled as an original image distorted by a T2 distortion filter. If the frequency response of the T2 distortion filter is known, inverse filtering can recover the original image from the acquired data. That is,

Detailed Description Text (22):
The main problem with the inverse filter is its sensitivity to measurement noise. When the acquired data S(k.sub.x,k.sub.y) is in the form of the T2 distorted signal plus noise, which is the case in any practical imaging system, we have

Detailed Description Text (23):
where N(k.sub.x,k.sub.y) is the measurement noise component. Then the inverse filter will give ##EQU3## Since the noise component N(k.sub.x,k.sub.y) is not affected by the T2 filter and H(k.sub.x,k.sub.y).ltoreq.1, we have N (k.sub.x,k.sub.y).gtoreq.N(k.sub.x,k.sub.y). The measurement noise will be amplified in this T2 correction process and the resulting images are often not acceptable.

Detailed Description Text (24):
The Wiener algorithm provides a better solution to the T2 correction problem in a noisy environment. Let the signal s(n.sub.x,n.sub.y) and the noise n

(n.sub.x,n.sub.y) be arbitrary, zero mean, random sequences, respectively. If the acquired sequence s(n.sub.x,n.sub.y) is modeled as

Detailed Description Text (25):
where h(n.sub.x,n.sub.y) is the distortion filter and ** denotes 2-D convolution, then the best linear estimate s(n.sub.x,n.sub.y) for the original signal s (n.sub.x,n.sub.y) can be obtained from the distorted data s(n.sub.x,n.sub.y) using a Wiener filter w(n.sub.x,n.sub.y) in the sense that the mean square error between the original signal s(n.sub.x,n.sub.y) and the estimated signal s(n.sub.x,n.sub.y) defined by

Detailed Description Text (28):
In MR imaging, the power spectrums of the signal and of the noise are not known and thus have to be estimated from the acquired data. Since S(k.sub.x,k.sub.y) is the inverse Fourier transform of the object, it usually has smaller values at higher spatial frequencies and thus the noise is often dominant in these frequencies. This will be especially true for hybrid fast MR imaging using a low-pass scheme since longer echo times further attenuate these high frequency signal components. Therefore, the average noise power P.sub.nn can be rather accurately estimated from the data S(k.sub.x,k.sub.y) on a support S.sub.n in the high spatial frequencies from the latest echo in the echo train using the standard formula ##EQU7## where N.sub.p is the total data number used for the P.sub.nn estimation on the data support S.sub.n. Since the measurement noise N(k.sub.x,k.sub.y) is often assumed to be white, we have the same noise power P.sub.nn everywhere in K-space. In practice, some DC component may exist. In this case, any such DC should be removed before making the estimation by equation (19).

Detailed Description Text (29):
Then the periodogram spectral estimation is used to obtain an estimate of the spectrum of the image data. [S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988] That is, we use the local data power P.sub.dd (k.sub.x,k.sub.y) defined by

Detailed Description Text (30):
to replace the signal power spectrum P.sub.ss (k.sub.x,k.sub.y) in equation (16) and thus obtain the frequency response of K-space Wiener T2 correction filter is ##EQU8## Since the frequency response H(k.sub.x,k.sub.y) of the T2 distortion filter is real in our discussion, the conjugate operation (*) and the absolute value operation (.vertline...vertline.) have been dropped in equation (21). The signal power can also be estimated by

Detailed Description Text (31):
This can be used in equation (21)instead of P.sub.dd (k.sub.x,k.sub.y) and will, in general, give stronger noise suppression. Once the frequency response of the Wiener filter is determined, the restored K-space image data can be obtained as ##EQU9##

Detailed Description Text (32):
One approach to determine the frequency response function H(k.sub.x,k.sub.y) from the acquired data is described below and from this the Wiener filter equation (21) can be completely specified for the image data restoration.

Detailed Description Text (33):
Since the frequency response H(k.sub.x,k.sub.y) of the T2 distortion filter depends on the spatial distribution and spin density of T2 values within the object under study, one cannot determine the exact frequency response of the T2 filter with just one data acquisition. Nevertheless, one approach for estimating H(k.sub.x,k.sub.y) from the acquired data based on the average amplitude symmetry constraint of the spin echo signal is set forth below.

Detailed Description Text (34):

It is well known from MR spectroscopy that the time-domain echo signal is symmetric around its peak if the T2 value is much longer than the echo observation time. In MR imaging, when the same TE is used for all phase encode (k.sub.y) values, ideally, the K-space data S(129+k.sub.x,129+k.sub.y) and S(129-k.sub.x,129-k.sub.y) should be conjugate symmetric about the peak signal position (k.sub.x =129, k.sub.y =129) in a 256.times.256 data matrix and have the same amplitude values if the T2 values of the object are much longer than the sampling window width in the frequency encoding (read-out, k.sub.x) direction. [D. A. Feinberg, J. D. Hale, J. C. Watts, L. Kaufman and A. Mark, "Halving MR Imaging Time by Conjugation: Demonstration at 3.5 kG." Radiology 164, pp. 527-31, 1986] When the T2 distortion effect exists, data amplitude distributions are biased. Therefore, one can estimate a global T2 value from the acquired data using this symmetric constraint.

Detailed Description Text (35):
FIG. 6 is shown a prone of the K-space data .vertline.S(k.sub.x, 129).vertline. for a rectangular object. It is symmetric about and peaked at k.sub.x =129. The curve on the top part of FIG. 6 is the exponential function e.sup.-2 t/T2 with T2=100 ms and the sampling window width T.sub.x =10 ms. If there is no noise and there is only a global T2 effect, any two values of .vertline.S(k.sub.x,129).vertline. from the two points symmetric about the peak would provide an estimate of the global T2 value. But there are many factors, including measurement noise and phase shifts, which affect the amplitude symmetry property. Therefore, one may first calculate the sums of the acquired signal amplitude .vertline.S(k.sub.x,129).vertline. over tow equal time spans symmetric about the echo peak. For example, if S(k.sub.x, 129) has N.sub.x samples in the k.sub.x direction with the sampling window time T.sub.x, we calculate the average amplitude A1 and A2 over a fraction of T.sub.x as ##EQU10## where N.sub.T2,x is the number of data samples in the T2 estimate window T.sub.w. An average T2 value can be calculated from these two amplitude values as

Detailed Description Text (36):
where T.sub.p is the period of the data sampling in the k.sub.x direction and thus T.sub.x =N.sub.x T.sub.p. From equation (1), it is known that ##EQU11## and thus the echo signal S(k.sub.x,129) is composed of contributions from all elements s (n.sub.x,n.sub.y) of the object. Therefore, (25) provides only a global T2 estimate of the object.

Detailed Description Text (37):
The derivation of equation (25) is based on an assumption that the signal has a flat Fourier spectrum over the T2 estimate window T.sub.w. For real signal spectrums with arbitrary shapes, the estimated global T2 will deviate from the true global T2 value and the difference between the true T2 value and estimated T2 value is affected by the width T.sub.w and the shape of the signal spectrum S (k.sub.x,k.sub.y). When there is no noise and no phase error, a shorter estimate window generates a smaller estimate difference (bias) from the true T2 value. When there are noise and other disturbing factors, the estimate window cannot be too short. It is the summation (low-pass filtering) process that reduces the disturbing effects such as noise and phase shifts, and helps obtain a stable T2 estimation. This is a typical tradeoff between the bias and the variance in the estimation problem [L. L. Scharf, Statistical Signal Processing: Detection, Estimation, and Time Series Analysis, Addison-Wesley Publishing Company, 1990] In practice, the T2 estimate window size T.sub.w should be adjusted according to the MR imaging conditions.

Detailed Description Text (40):
A1 and A2 could also be calculated over a wider window in k.sub.y direction such that ##EQU13## where N.sub.T2,y determines the width (2N.sub.T2,y +1) of the T2 estimate window in the k.sub.y direction. When N.sub.T2,y =0, equation (29) reduces to equation (27). After a global T2 value has been estimated, the frequency response H(k.sub.x,k.sub.y) of the T2 distortion filter can be determined using equation (7), according to the MR sequence specification of echo times TE(k.sub.y)

and the data sampling period T.sub.p.

Detailed Description Text (41):
The approach discussed above makes the estimation of H(k.sub.x,k.sub.y) from the acquired image data directly and thus does not require extra data acquisition. But this approach can only estimate the amplitude distortion function of the T2 filter and therefore the resulting Wiener filter based on these estimates can only reduce the amplitude distortion caused by the T2 effect.

Detailed Description Text (42):
Linear prediction techniques have been successfully used for time-series analysis, high resolution spectrum estimation, speech and image signal coding, and many other applications [S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988] The technique has also been used for MR data extrapolation to reduce the truncation artifacts and improve spatial resolution [M. R. Smith, S. T. Nichols, R. M. Henkelman and M. L. Wood, "Application of Autoregressive Moving Average Parametric Modeling in Magnetic Resonance Image Reconstruction," IEEE Trans. Med. lmag., Vol. 5, pp. 132-39, 1986; J. E Martin and C. F. Tirendi, "Modified Linear Prediction Modeling in Magnetic Resonance Imaging," J. Magn. Reson., Vol. 82, pp. 392-99, 1989; E. M. Haacke, Z. Liang and S. H. Izen, "Super Resolution Reconstruction Through Object Modeling and Parameter Estimation," IEEE Trans. ASSP., Vol. 37, pp. 592-95, 1989]. The application of linear prediction to local T2 correction of both amplitude and phase distortions is discussed below.

Detailed Description Text (45):
Assume that the global T2 corrected MR data by the Wiener filter are represented by a 2-D function S(k.sub.x,k.sub.y) where k.sub.x is the index of sampling points in the frequency encode direction and k.sub.y is the index of sampling points in the phase encode direction. First, the inverse Fourier transform of the time-domain data S(k.sub.x,k.sub.y) is taken with respect to k.sub.x to obtain the hybrid-domain data set X(n.sub.x,k.sub.y) and then the data X(n.sub.x,k.sub.y) is considered for each n.sub.x value. Given the discontinuity positions in the frequency response of T2 distortion filters, the linear prediction is used to produce data samples across these positions. For example, if the T2 distortion filter is as shown in FIG. 2, a total of 31 data samples from X(n.sub.x,130) to X (n.sub.x,160) can be used to predict data samples on the other side of the discontinuity position by forward linear prediction, starting from k.sub.y =161. The prediction is performed from data samples of lower frequency to data samples of higher frequency since low frequency data samples usually have higher signal-to-noise ratio and therefore the resulting predicted data have lower prediction errors.

Detailed Description Text (47):
This local phase distortion factor is then used to make a local zero-order phase correction by multiplying all the old data samples of the second echo from k.sub.y =161 to k.sub.y =192. The resulting phase corrected data are denoted as X (n.sub.x,k.sub.y). That is

Detailed Description Text (48):
The estimated local phase distortion factor .PSI.(n.sub.x) from different column data X(n.sub.x,k.sub.y) can be first low-pass filtered and the resulting smoothed phase estimates are then used in equation (34) for phase correction. This enhances the stability of local phase correction.

Detailed Description Text (49):
Since the prediction usually has quite precise results for data samples close to the starting position (k.sub.y =161) in the echo train boundary and becomes less accurate as k.sub.y moves away from the boundary, predicted data X(nx,k.sub.y) are combined with zero-order phase corrected data x(n.sub.x,k.sub.y) as a weighted

average over a data merging band to obtain local amplitude and phase corrected data X(n.sub.x,k.sub.y). That is,

Detailed Description Text (50):
where N.sub.m is the width of the data merging band. Two weighting factors w.sub.1 (k.sub.y) and w.sub.2 (k.sub.y) are some monotonously decreasing and increasing functions of k.sub.y, respectively. They are related as w.sub.1 (k.sub.y)+w.sub.2 (k.sub.y)=1.0 for all k.sub.y. The use of these weighting functions are intended to ensure a smooth transition from the predicted data to the originally observed data. In this study, w.sub.1 (k.sub.y) and w.sub.2 (k.sub.y) are chosen as linear functions of k.sub.y and thus ##EQU16## Although other weighting functions such as quadratic and exponential functions have been tried to make different weightings on the two data sets, it has been found that they do not make a significant performance difference. The determination of the data merging band N.sub.m is affected by several considerations. A wider merging band usually provides stronger ringing artifact reduction but it also results in bigger changes in the original signal spectrum. Also, its value is limited by the data number in each echo group. A band around ten data samples gives good results for four-echo HI images of 256 phase encoding projections.

Detailed Description Text (51):
A similar local amplitude and phase correction is performed around other discontinuity positions in the data set X(n.sub.x,k.sub.y) except that for the discontinuity positions of k.sub.y <129, a backward linear prediction is used to estimate data samples across these positions (based also on the data samples with lower frequencies). Finally, the inverse Fourier transform of the hybrid domain data X(n.sub.x,k.sub.y) is taken with respect to k.sub.y to reconstruct the HI image. As shown by the subsequently discussed examples, this processing effectively reduces effects from amplitude and phase discontinuities in the frequency response of the effective T2 distortion filter and therefore reduces ringing artifacts in the reconstructed HI image.

Detailed Description Text (53):
Because the phase error has not been corrected, it is noted that image resolution loss and ringing artifacts still exist. If local amplitude and phase correction is applied to the distorted image data directly, a local corrected image is obtained as shown in FIG. 7C with improved resolution and reduced ringing artifacts. In this case, 30 data samples were used to calculate the autocorrelation function (ACF) with a biased ACF estimator [L. B. Jackson, Digital Filters and Signal Processing, Second Edition, Kluwer Academic Publishers, 1989] For the predictor order selection, there are many techniques available [S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988] The model order selection rule N/3.ltoreq.p.ltoreq.N/2 by Ulrych and Bishop [T. J. Ulrych and T. N. Bishop, "Maximum Entropy Spectral Analysis and Autoregressive Decomposition," Rev. Geophys. Space Phys., Vol. 13, pp. 183--200, 1975] where N is the number of data samples used for the ACF estimation was used here. Therefore, the order p of the linear predictor for this case has been chosen to be 12. The predictor coefficients are determined with the Levinson-Durbin algorithm. Then new data samples can be predicted using equations (30) or (31) based on known data samples and previously predicted data samples. The estimated local phase distortion using equation (33) from the forward predicted data X(161) and the old data sample X(161) is 97.15.degree. and the estimated local phase distortion is 92.54.degree. from the backward predicted data X(96) and the old data X(96). The phase estimate error is less than 10%. As described above, these phase estimates are used to make the zero-order phase correction on data of k.sub.y .gtoreq.161 and k.sub.y .ltoreq.96, respectively. Then the phase corrected data are merged with the estimated data to form a new data set for final image reconstruction. A data merging band of ten data samples has been used for all discontinuity positions in FIG. 4.

Detailed Description Text (54):
Finally, the local amplitude and phase correction technique is employed on data processed by the global T2 amplitude correction. The resulting corrected HI image by this two-stage procedure is given in FIG. 7D. It is clear from these figures that the original HI image degradation caused by amplitude attenuation, resolution loss and ringing artifacts has been effectively improved by this combined global and local data processing technique (except near the sharp edges of the objects).

Detailed Description Text (55):
The proposed method has been applied to an image of a phantom acquired in a low-field-strength (0.064 T) permanent magnet imager (ACCESS.TM. Toshiba-America MRI Inc.). The phantom object is a rectangular container full of mineral oil with a T2 value of about 100 ms.

Detailed Description Text (56):
The asymmetric Fourier imaging (AFI) approach has been employed to further reduce the imaging time. A total of 144 phase encode projections has been acquired by 36 data acquisition shots using a four-echo band-pass sequence with echo times TE(i) =10.i ms for i=1,2,3,4. The data is then conjugated to a full image size of 256.times.256 pixels. The resulting T2 filter has a frequency response similar to that shown in FIG. 4 and phase encode is in the vertical direction. Because of amplitude and phase distortions introduced by the T2 distortion filter, the reconstructed image has serious tinging artifacts along the phase encoding direction as evidenced in FIG. 8A. For demonstration purposes, one of the worst cases of ringing artifacts is illustrated.

Detailed Description Text (57):
With a global T2 estimate window of N.sub.T2,x =64, N.sub.T2,y =2 and the root factor .alpha.=0.5, the global T2 value estimated using equations (27) and (28) is 127 ms. This estimated T2 value is used in a Wiener filter for global amplitude correction on the originally acquired 144 projections. Now, 21 data samples are used for the ACF estimate and the predictor order is ten. For the AFI data case, both the global and the local corrections should be performed before conjugation and this will save computations needed for the data correction. For the given AFI data of 144 phase projections, the saving in the global T2 amplitude correction is about 44% compared with the full size data of 256 phase projections. For local corrections, the saving is 50% since only three instead of six discontinuities need to be processed. T2 corrected data are conjugated to the full image size and the improved image as shown in FIG. 8B is obtained which has much less ringing artifact. The standard deviation of the noise has been reduced form 144 to 55 by the Wiener filtering.

Detailed Description Text (58):
As a further example, one image from a sequence with both sin echo and gradient echo has been tested. The sequence consists of three echoes in the echo train, one spin echo and two gradient echoes. The spin echo has an echo time 25 ms and generates the signal components S(k.sub.x,k.sub.y) for 86.ltoreq.k.sub.y .ltoreq.170. The first graident ehco has an echo time 19 ms and generates signal components S(k.sub.x,k.sub.y) for 1.ltoreq.k.sub.y .ltoreq.85. The second gradient echo has an echo time 32 ms, generating the remaining signal components. The image was acquired in a high-field-strength (1.5 T) MR image (Toshiba MRT200/FXIII). Similar to the fast spin echo case, because of T2 and T2* effects, the tinging artifacts in the reconstructed image are obvious as shown in FIG. 9A, which seriously degrade the image quality. In this image, phase encoding is in the horizontal direction.

Detailed Description Text (59):
Although the echo times given above seem to define a ramp T2 filter, the actual T2 filter has a low-pass shape since the dephasing effects (T2* effects) caused by field inhomogeneity results in faster decay for the gradient echo signal components

than from the T2 effect. In this case, the T2 amplitude attenuation factors A.sub.g1 for the first gradient echo and A.sub.g2 for the second gradient echo are estimated using signals from three extra zero phase encoding projections in the end of the sequence. They are determined by ##EQU17## where S.sub.s (k.sub.x,129) is the zero-phase encoding signal from the standard spin echo, S.sub.g1 (k.sub.x,129) and S.sub.g2 (k.sub.x,129) are the zero-phase encoding signals from the first and the second gradient echoes, respectively. For the given image, estimated attenuation factors are A.sub.g1 =0.84 and A.sub.g2 =0.72 with respect to the spin echo signal. These attenuation factors were used in the Wiener filter to make a global amplitude correction. This estimation method for global T2 amplitude attenuation factors can also be used for the FSE case. Then, the Wiener filter and the linear prediction are applied to this image and the resulting image after the two-stage processing is shown in FIG. 9B (which has imroved resolution and fewer tinging artifacts). The standard deviation of noise in the jost homogeneous tissue region has also been reduced from 149 to 118.

Detailed Description Text (60):
In the above experiments, the Wiener filter and the linear prediction technique have been used together to effectively improve HI image quality. For some applications, the Wiener filter can be used only to perform partial global T2 amplitude correction. For example, the sequence with a band-pass frequency response as shown in FIG. 4 is often used to emphasize signal components with middle range frequencies and obtain an edge enhance dimage. In this case, for preserving the desired objective, a global Wiener T2 amplitude correction can be employed only for signal components from the third and fourth echoes to further increase high frequency components (leaving the low frequency components from the second echo untouched). Then the linear prediction technique is used to reduce tinging artifacts.

Detailed Description Text (61):
The combined use of a Wiener filter and a linear prediction technique for T2 restoration and noise suppression has been presented. The effectiveness of this method in improving HI image quality has been illustrated using some experimental results in which only one estimated global T2 value (and therefore a simple frequency response function H(k.sub.x,k.sub.y) of T2 filter) is available for the T2 amplitude restoration. This simplified model may result in under and/or over compensations for certain objects in a given image. Methods for more accurate estimate of T2 values and distributions are required to obtain better T2 correction. For local amlpitude and phase correction, the simple linear prediction has been used for computation simplification. Other modeling techniques such as autoregressive moving average (ARMA) [M. R. Smith, S. T. Nichols, R. M. Henkelman and M. L. Wood, "Application of Autoregressive Moving Average Parametric Modeling in Magnetic Resonance Image Reconstruction," IEEE Trans. Med. Imag., Vol. 5, pp. 132-39, 1986; S. M. Kay, Modern Spectral Estimation: Theory & Application, Prentice Hall, Englewood Cliffs, N.J. 07632, 1988; L. B. Jackson, Digital Filters and Signal Processing, Second Edition, Kluwer Academic Publishers, 1989] and nonlinear prediction with neural network [H. Yan and J. Mao, "Data Truncation Artifcat Reduction in MR Imaging Using a Multilayer Neural Network," IEEE Trans. Med. Imag., Vol. 12, pp. 73-77, 1993] can be employed to improve the results.

Detailed Description Text (62):
As indicated above, this invention may be implemented by suitably programming the image data processor of a conventional MRI system so as to implement the above-described two-stage T2 correction (global, then local) process. The program may be provided as optional additional subroutines if desired or incorporated as permanently used non-optional added data processing routines. In any such case, the programs may be conventionally written by those skilled in the art to straightforwardly implement the above-described processes. Accordingly, no further detailed description of such computer programs is believed to be necessary.

Detailed Description Paragraph Equation (2):
S(k.sub.x,k.sub.y)=N(H)(1-e.sup.-TR/T1)e.sup.-(TS(k.sub.x)+TE)/T2S(k.sub .x, k.sub.y) (3)

Detailed Description Paragraph Equation (3):
S(k.sub.x,k.sub.y)=N(H)(1-e.sup.-TR/T1)e.sup.-(TS(k.sub.x)+TE(k.sub.y))/ T2S (k.sub.x,k.sub.y). (4)

Detailed Description Paragraph Equation (7):
H(k.sub.x,k.sub.y)=e.sup.-(TS(k.sbsp.x.sup.)/T2 e.sup.-TE(k.sbsp.y.sup.)/T2 =H.sub.x (k.sub.x)H.sub.y (k.sub.y) (8)

Detailed Description Paragraph Equation (8):
H.sub.x (k.sub.x)=e.sup.-TS(k.sbsp.x.sup.)/T2, H.sub.y (k.sub.y)=e.sup.-TE (k.sbsp.y.sup.)/T2 (9)

Detailed Description Paragraph Equation (20):
X(n.sub.x,k.sub.y)=w.sub.1 (k.sub.y)X(n.sub.x,k.sub.y)+w.sub.2 (k.sub.y)X (n.sub.x,k.sub.y) for 161.ltoreq.k.sub.y .ltoreq.(161+N.sub.m) (35)

CLAIMS:

1. A method for enhancing MRI data acquired at different NMR echo times to compensate for T2 and noise distortions, said method comprising the steps of:

estimating a global T2 value from acquired MRI data; and

applying a Wiener filter to said acquired data in the spatial frequency domain to effect a global T2 amplitude correction and noise suppression using said estimated global T2 value.

3. A method as in claim 1 wherein said Wiener filter approximates the inverse of an effective T2 distortion filter that is determined to have produced an asymmetry in the acquired MRI data.

4. A method as in claim 1 further comprising the steps of:

defining linear predictions of estimated local amplitude and phase for one-dimensional Fourier transformations of the globally corrected MRI data in the spatial frequency domain, and

utilizing said linear predictions to make local amplitude and phase corrections to such data during MR image reconstruction processing.

5. A method as in claim 2 further comprising the steps of:

defining linear predictions of estimated local amplitude and phase for one-dimensional Fourier transformations of the globally corrected MRI data in the spatial frequency domain, and

utilizing said linear predictions to make local amplitude and phase corrections to such data during MR image reconstruction processing.

6. A method as in claim 3 further comprising the steps of:

defining linear predictions of estimated local amplitude and phase for one-dimensional Fourier transformations of the globally corrected MRI data in the spatial frequency domain, and

utilizing said linear predictions to make local amplitude and phase corrections to

such data during MR image reconstruction processing.

7. A method for enhancing MRI data acquired at different NMR echo times to compensate for T2 and noise distortions, said method comprising the steps of:

(a) estimating a global T2 value for an object being imaged from acquired MRI data based on the average amplitude symmetry constraint of an NMR spin echo signal;

(b) applying a Wiener filter to make a global T2 amplitude correction and noise suppression in said acquired MRI data in the spatial frequency domain;

(c) Fourier transforming the thus corrected data in the NMR signal read out direction to obtain a globally corrected hybrid domain MRI signal set;

(d) utilizing a linear prediction technique to provide estimates of local NMR signal amplitude and phase, and

(e) utilizing said estimates of local NMR signal amplitude and phase to make local amplitude and phase corrections to the data resulting from step (c).

8. Apparatus for enhancing MRI data acquired at different NMR echo times to compensate for T2 and noise distortions, said apparatus comprising:

means for estimating a global T2 value from acquired MRI data; and

means for applying a Wiener filter to said acquired data in the spatial frequency domain to effect a global T2 amplitude correction and noise suppression using said estimated global T2 value.

10. Apparatus as in claim 8 wherein said Wiener filter approximates the inverse of an effective T2 distortion filter that is determined to have produced an asymmetry in the acquired MRI data.

11. Apparatus as in claim 8 further comprising:

means for linear predictions of estimated local amplitude and phase for one-dimensional Fourier-transformations of the globally corrected MRI data in the spatial frequency domain, and

means for utilizing said linear predictions to make local amplitude and phase corrections to such data during MR image reconstruction processing.

12. Apparatus as in claim 9 further comprising:

means for linear predictions of estimated local amplitude and phase for one-dimensional Fourier-transformations of the globally corrected MRI data in the spatial frequency domain, and

means for utilizing said linear predictions to make local amplitude and phase corrections to such data during MR image reconstruction processing.

13. Apparatus as in claim 10 further comprising:

means for linear predictions of estimated local amplitude and phase for one-dimensional Fourier-transformations of the globally corrected MRI data in the spatial frequency domain, and

means for utilizing said linear predictions to make local amplitude and phase corrections to such data during MR image reconstruction processing.

14. Apparatus for enhancing MRI data acquired at different NMR echo times to compensate for T2 and noise distortions, said apparatus comprising:

(a) means for estimating a global T2 value for an object being imaged from acquired MRI data based on the average amplitude symmetry constraint of an NMR spin echo signal;

(b) means for applying a Wiener filter to make a global T2 amplitude correction and noise suppression in said acquired MRI data in the spatial frequency domain;

(c) means for Fourier transforming the thus corrected data in the NMR signal read out direction to obtain a globally corrected hybrid domain MRI signal set;

(d) means for utilizing a linear prediction technique to provide estimates of local NMR signal amplitude and phase, and

(e) means for utilizing said estimates of local NMR signal amplitude and phase to make local amplitude and phase corrections to the data resulting from the means for Fourier transforming.

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Lee; David C.	The Peak			HK
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Spielman; Steven R.	Berkeley	CA	94703	

US-CL-CURRENT: 369/47.19; 369/124.12

Full | Title | Citation | Front | Review | Classification | Date | Reference | | | Claims | KWIC | Draw D

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Term	Documents
K-SPACE	2184
K-SPACES	36
KSPACE	51
KSPACES	2
"K SPACE"	0
KX	5842
KXES	20
KY	50047
KIES	989
KYS	253
KZ	661216
(L31 AND (((CORRECT$4 OR COMPENSAT$4) WITH (ACQUIR$4 OR OBTAIN$4 OR COLLECT$4 OR GATHER$4) WITH (K-SPACE OR KSPACE OR "K	

SPACE" OR "KX" OR "KY" OR "KZ" OR "K.SUB.X" OR "K.SUB.Y" OR "K.SUB.Z" OR RAW) WITH (DATA OR ECHO OR SIGNAL)) WITH ((((AMPLITUDE OR INTENSITY OR MAGNITUDE OR PEEK OR PEAK) WITH (MODULATION OR MODULATING OR MODULATE OR MODIFY OR MODIFYING OR MODIFIED OR CHANG$4 OR ADJUST$4 OR ALTER$3 OR VARY OR VARIED OR VARYING OR VARIABLE)) OR "AM")))).PGPB,USPT,USOC,EPAB,JPAB,DWPI,TDBD.	4

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